



Marriage of CT and MRI for vulnerable plaque characterization

“By omni-tomography, we envision that the next stage of biomedical imaging will be the grand fusion of many tomographic modalities into a single gantry (‘all in one’) for simultaneous data acquisition of many complementary features (‘all at once’).”

KEYWORDS: cardiac imaging ■ compressive sensing ■ CT ■ CT–MRI ■ interior tomography ■ MRI ■ omni-tomography ■ vulnerable plaque characterization

While CT, MRI, PET, SPECT and ultrasound have their well-defined roles for medical imaging, over the past decade multimodality systems, such as PET–CT and PET–MRI scanners, have become increasingly popular for sequential or contemporaneous data acquisition. However, these paired modalities impose limitations that may compromise our understanding of physiological processes relative to fine details and rapid changes driven by a beating heart. Not all pairs of tomographic scanners can be engineered closely together or within a single gantry, given the bulkiness of individual systems, such as CT and MRI scanners, and the conflict in imaging physics, such as rotating metallic parts for CT and magnetic fields for MRI.

Recently, our group has been performing pilot studies on what is called omni-tomography [1]. This strategy is conceptualized based on interior tomography developed over the past 5 years [2]. By omni-tomography, we envision that the next stage of biomedical imaging will be the grand fusion of many tomographic modalities into a single gantry (‘all in one’) for simultaneous data acquisition of many complementary features (‘all at once’).

Now, novel multifunctional probes, multi-/coupled-physics modeling, high-technology engineering and advanced image reconstruction, especially interior tomography and compressive sensing, present new opportunities to peek into living biological systems noninvasively, systematically and concurrently. Building blocks are either available or emerging for the initial development of omni-tomography. As an inspiring case of omni-tomography, here we propose the first CT–MRI scanner for vulnerable plaque characterization, which is described as the ‘holy grail’ of cardiology [101].

With the CT–MRI scanner, CT and MRI scans can be seamlessly merged for spatiotemporal registration, extendable to include other imaging modalities. CT provides structural definition in snapshot. MRI reveals blood flow, soft tissue contrast and functional, cellular and molecular features. Neither CT nor MRI individually could cover all biomarkers of vulnerable plaques, including cap thickness, lipid-core size, stenosis, calcification, hemorrhage, elasticity, inflammation, endothelial status, oxidative stress, platelet aggregation, fibrin deposition, enzyme activity, microbial antigens, apoptosis, angiogenesis and so on. It would be exciting to have all of these features in high spatial, contrast, temporal resolution within a common coordinate system. Even if MRI of vulnerable plaques could have sufficient resolution and speed, retrospective image registration between CT and MRI is not a desirable alternative, owing to registration errors due to nonrepeatable contrast dynamics, organ motion and deformation, MRI-induced geometric distortion and signal nonlinearity, as well as inconsistent contrast mechanisms between CT and MRI.

Our proposed CT–MRI scanner is a major refinement of our initial concept outlined in [1]. The key idea is to let CT and MRI focus on a relatively small region of interest (ROI). Hence, each imaging modality can be simplified and parallelized. They can then be accommodated within a single gantry without physical interference.

Traditional CT methods cannot exactly reconstruct an interior ROI solely from truncated projections along x-rays through the ROI. This interior problem has been studied for decades, and the fact that precise reconstruction could not be obtained from local data contributed

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to the longstanding CT architectures whereby detectors always fully cover a transverse slice. Now, interior tomography allows for theoretically exact reconstruction over an ROI from purely local data aided by practical prior knowledge. More importantly, interior tomography has been extended as a general imaging principle and developed for other modalities such as MRI [1], SPECT [3] and others. Thanks to interior tomography, data acquisition modules of different types can be made more compact or sparse to provide space and reduce cost for omni-tomography.

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Our proposed CT–MRI system is mainly based on the generalized interior tomography principle and two pieces of prior art. First, an existing open MRI architecture gives open space between two donut-shaped magnets for CT. Second, the work on combination of x-ray imaging and MRI shows the compatibility of x-ray imaging and MRI [4]. The CT subsystem consists of nine x-ray tubes, each of which is paired with a detector array. This allows instantaneous data acquisition of nine views targeting a heart. Nine views could be enough in special cases, but are generally insufficient for ROI reconstruction of diagnostic quality. Hence, the multisource data acquisition assembly can be rotated (or the patient table can be properly tilted) for three or five sets of nine views. Each data acquisition session can be synchronized to the ECG signal with breath-holding. In an alternative design, 45 or more carbon nanotube x-ray focal spots can be distributed along half a circle or a full circle for rapidly multiplexed acquisition in nine-view groups.

The x-ray tubes in the proposed CT–MRI system will be under a significant magnetic field (~0.5 T with the current design for a 1.5 T background field in the field of view). Hence, the electron beams in the x-ray tubes will interact with the magnetic field. The magnetic field can affect the trajectories of both the primary electrons (i.e., the electron from the cathode) and the backscattered electrons (i.e., the electrons backscattered from the anode). The primary electrons can be deflected by the magnetic field [4]. This deflection problem is more severe when the magnetic field is oblique to the cathode–anode electric field in the x-ray tube. Even if the magnetic field is parallel to the cathode–anode electric field, electron

beam defocusing remains problematic because electrons emitted from a thermionic cathode are omnidirectional and have a distribution of initial velocities. When the accelerated primary electrons strike the anode, a significant portion of those electrons will be backscattered. For an x-ray tube working at 65 kV, more than 50% of the electrons incident on the target are backscattered [5]. Without a magnetic field, only a small fraction of these backscattered electrons will return to the anode. However, when a strong magnetic field exists, the backscattered electrons will have an elevated probability of striking the anode again, resulting in an increased total x-ray output, slightly softening the x-ray spectrum and enlarging the focal spot size [5].

In our proposed CT–MRI system, the x-ray tubes will work in the static main magnetic field from the main magnets without complication of the magnetic fields from the gradient and radio frequency coils. The main magnetic field is parallel to the longitudinal direction of the CT–MRI system. Using a reflection-type anode at a suitable takeoff angle, the cathode–anode axes in all the x-ray tubes can be made parallel to the direction of the magnetic field. Our simulation study indicates that a 0.5 T static magnetic field is expected at the x-ray tube position. To prevent the backscattered electrons from returning to the focal spot area, active shielding can be used to minimize the magnetic field near the focal spot [6].

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The MRI subsystem uses a highly sophisticated superconducting electromagnet technology. The performance of the paired superconducting magnets is critical to the image quality. The static field produced by the magnets in the field of view is relaxed to be much smaller than the counterpart for a conventional MRI scanner, and needs to be strong and only nearly perfect (a few ppm variation) throughout a cardiac ROI. This represents a major simplification relative to the requirements for a much larger conventional field of view. The design of the split architecture with a large central gap (>40 cm) is challenging, demanding the full consideration of the electromagnetic, mechanical and thermal properties of all the involved components.

Our superconducting MRI magnet has a two-layer coil configuration. The first layer with a small radius (50 cm) provides a primary magnetic field in the field of view, while the second layer

with a large radius (80 cm) is mainly for shielding the stray magnetic field within the domain of interest (8 m in length and 5 m in radius). The magnet design offers a 1.5 T field of view of an appropriate diameter (25 cm), which defines a spherical field of view. The field uniformity inside the spherical field of view is 1 ppm (peak–peak). This magnet design allows the integration of CT in the gap between the pair of magnets. During the design, a current density pattern was first derived with a simplified linear field–source relationship. Then, wire layouts were determined using a nonlinear optimization scheme [7]. A quench simulation study was conducted to balance imaging performance and system cost. Since engineering errors unavoidably compromise field uniformity, a shimming technique must be used for field correction. In this project, a robust field-based passive-shimming algorithm was used to control the field uniformity and harmonics [8]. Moreover, current-based active shimming techniques were employed to improve the field homogeneity throughout the spherical field of view. Three orthogonal gradient coils were designed for the split magnet bore to offer linear gradient fields. In particular, paired saddle coils were used for the x- and y-gradient coils. A Maxwell pair served as the z-gradient coil. More sophisticated coil patterns are possible for

advanced MRI applications [9]. To transmit and acquire radio frequency signals, array coils can be arranged around the spherical field of view [10].

Although the above CT–MRI system design is based on physical and engineering principles, construction of a real system is not an easy task, since it is significantly more complicated than any multimodality system on the market. We could start with proof-of-concept prototypes to show unique utilities, and then gradually move to product-ready platforms. For our ultimate goal of vulnerable plaque characterization, we would probably need 50 μm spatial resolution, 20 ms temporal resolution and other cutting-edge quality indices. These are far from immediately reachable but do seem unprecedented research opportunities.

Financial & competing interests disclosure

G Wang and H Gao have a pending patent (US Patent Application 20120265050, filed on 4 April 2012). The authors have a provisional patent on the CT–MRI scanner.

The authors have no other relevant affiliations or financial involvement with any organization or entity with a financial interest in or financial conflict with the subject matter or materials discussed in the manuscript apart from those disclosed.

No writing assistance was utilized in the production of this manuscript.

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