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Positron Emission Tomography/Magnetic Resonance Imaging of Gastrointestinal Cancers

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As an integrated system, hybrid positron emission tomography/magnetic resonance imaging (PET/MRI) is able to provide simultaneously complementary high-resolution anatomic, molecular, and functional information, allowing comprehensive cancer phenotyping in a single imaging examination. In addition to an improved patient experience by combining 2 separate imaging examinations and streamlining the patient pathway, the superior soft tissue contrast resolution of MRI and the ability to acquire multiparametric MRI data is advantageous over computed tomography. For gastrointestinal cancers, this would improve tumor staging, assessment of neoadjuvant response, and of the likelihood of a complete (R0) resection in comparison with positron emission tomography or computed tomography.
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The number of installed hybrid positron emission tomography or magnetic resonance imaging (PET/MRI) scanners has increased steadily since their clinical introduction in 2010. Current scanners combine a PET system with a 3-tesla (3 T) MRI, either as an integrated or sequential system.[1,2] In an integrated system, PET and MRI are colocated with the PET detectors embedded within the MRI system. In a sequential scanner, the MRI and PET are linked with a shuttle system, allowing the patient to be transferred directly from 1 system to another yet maintaining the same patient table position to facilitate co-registration of PET and MRI (Table).

Integration of PET with MRI has been challenging but eventually achieved through substantial engineering advances.[3] The MRI static B0 and radiofrequency B1 magnetic field and radiofrequency pulse would affect electron paths, cause heating, vibration, and interference with PET electronics. Likewise, PET components may affect the homogeneity of the MRI B0 field and radiofrequency detection. Integration has required the replacement of standard PET photomultiplier tubes with magnetic field-insensitive avalanche photodiodes or silicon photomultipliers, incorporation of additional shielding around the PET system and redesign of electronics.[3] MRI radiofrequency surface coils also have been designed to be "PET transparent" with minimal attenuation of emitted photons.

Integration has brought new challenges for PET attenuation correction, an essential part of PET imaging.[4] PET radionuclide decay results in the emission of 2 opposing 511 keV annihilation photons. Some photons are attenuated or scattered by body tissues before being detected causing signal loss proportional to the distance from the surface and regional tissue density. Non–attenuation-corrected PET data would generally underestimate tracer activity deep within the body. With a hybrid PET or computed tomography (CT) system, attenuation correction using CT-based X-ray attenuation, scaled to reflect the attenuation of higher energy 511 keV emission photons, enables accurate quantification.[5]

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Table Comparison of Current Hybrid PET/MRI Systems

Systems	Integrated		Sequential
Manufacturer	Siemens	General Electric	Philips
Scanner Bore	60 cm	60 cm	60 cm
MRI			
Field Strength	3 T	3 T	3 T
Gradients	45 mT/m	50 mT/m	40 or* 80 mT/ms
Slew rate	200 mT/m/s	150 mT/m/s	200 or* 100 mT/m/s
MRI Field-of-View (FOV)	50 x 50 × 45 cm	45 x 45 × 45 cm	50 x 50 × 45 cm
PET Detectors	Lutetium oxyorthosilicate avalanche photodiodes	Lutetium-based scintillator with silicon photomultipliers	Lutetium oxyorthosilicate photomultiplier detector modules
PET Axial FOV	25 cm	25 cm	18 cm
Simultaneous acquisition	Yes	Yes	No

*Enhanced mode.

With hybrid PET/MRI this is not possible, as MRI signal intensity is not based on tissue density as with CT, but on proton density and relaxation properties, requiring alternative methods.[4] In current systems, dedicated attenuation correction sequences may be acquired, most commonly a 2-point Dixon 3D gradient echo, deriving an attenuation map (μ map) based on the following 4 tissue types: background air, lungs, soft tissue, and fat. A limitation of such methodology is the classification of bone as soft tissue, which may lead to standardized uptake value underestimations of bone lesions by as much as 30%.[6-8] Ongoing work on ultra-short time-to-echo (TE) MRI sequences may improve bone classification at the expense of a longer acquisition time and possible artifacts with large field-of-view imaging (which is the norm for whole-body imaging).[9] Alternatives, including atlas-based methods and software-based artificial intelligence algorithms (eg, neural networks and fuzzy logic) to segment anatomical structures such as soft tissue and bone for attenuation correction, are also under consideration.[10]

Another issue for attenuation correction is that the MRI field-of-view is smaller than the PET field-of-view. This can affect MR attenuation correction if anatomical structures lie outside the field-of-view, for example, in obese patients. The arms may also be truncated by the MRI field-of-view. A current workaround in clinical systems is to iteratively extract the contours of the arms from PET data (maximum likelihood reconstruction of attenuation and activity[11]) and to use this information to complete the MR-based attenuation correction map of the body.

Currently, hybrid PET/MRI is evolving from being just a research tool to include clinical applications, as the high sensitivity of PET acts synergistically with the high contrast and spatial resolution of MRI. It is also possible to include locoregional MRI functional sequences, including diffusion-weighted MRI, dynamic contrast-enhanced MRI, and blood oxygenation level–dependent MRI to provide a more comprehensive evaluation of underlying tumor biology and physiology. This is particularly promising for imaging gastrointestinal cancers.

Nevertheless, a 3-T MRI system does bring some challenges for body imaging.[12] Although the signal-to-noise ratio increases approximately linearly with field strength, and thus an advantage of imaging at 3 T, the energy deposition would also increase as the square of field strengths. Specific absorption ratio limits are likely to be reached more often, typically restricting acquisition parameters, for example, lengthening the acquisition time.

Standing wave effects are also relevant at 3 T. The ^{1}H Larmor resonance frequency increases with field strength, for example, from 63.9 MHz at 1.5 T to 127.8 MHz at 3 T; consequently, the wavelength of the applied radiofrequency (in the medium water and thus "body tissue") is shortened, for example, from 52 cm at 1.5 T to 26 cm at 3 T in body tissue (vs 470-235 cm in the medium air). This 26 cm wavelength is similar to the dimensions of the body trunk in axial sections, and relevant as radiofrequency excitation is undertaken perpendicular to the body long axis. The shorter wavelength inside the body at 3 T results in a negative interference of superimposed radiofrequency waves emanating into the body from the surface. Near-complete cancellation of radiofrequency excitation may occur in central parts of the abdomen or pelvis. Further radiofrequency effects may also arise from induced eddy currents in conductive parts of the body. Both effects result in marked B1 field inhomogeneity (and local irregular flip angles) that are relevant to centrally located gastrointestinal cancers and quantitative functional MRI techniques and may be worse in patients with increased conductivity, for example, with ascites that acts as a conductive medium. Workarounds in clinical systems include improved coil designs, more flexible radiofrequency shimming, and draining of ascites before imaging.

Another issue is that the chemical shift effect is greater at 3 T. Chemical shift results from partial shielding of the outer magnetic field at the location of the nuclei by the electron sheath of the molecule. This is proportional to the field strength: the Larmor frequency difference between nuclei with different chemical bonds increases linearly, for example, the difference between water and methylene (in fatty acid and triglycerides) doubles from 220-440 Hz at 3 T. This is relevant to gastrointestinal imaging because of the fat or water interface at the bowel wall that may lead to chemical misregistration artifacts. This has to be considered when assessing tumor stage (T stage).

Finally, magnetic susceptibility effects also increase linearly with field strength and is 2 times greater at 3 T when compared