# Optimization of MR and PET image quality for breast imaging with the Biograph mMR

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**Introduction:** The specificity of MR in the detection of breast cancer has been shown to be significantly increased when combined with spatially fused $^{18}$F-fluorodeoxyglucose (FDG) PET images [1]. Several factors, however, limit the efficacy of spatially registering images acquired from standalone MR and PET systems. For instance, the breast is a highly deformable organ and current state of the art registration methods have limited accuracy even when matched patient positioning is used. Combined MR-PET scanning is expected to significantly improve spatial registration between the separate modalities and as such may improve patient care in the management of breast cancer. Since no breast coil has been developed for the Biograph mMR MR-PET (Siemens Healthcare), we aimed to optimize image quality for breast imaging for both the MR and PET components using a conventional MR breast coil.

**Materials and Methods:** Both MR and PET image quality was optimized by imaging a 1.9 L uniformly filled cylinder phantom (max diameter=12 cm) with a 4-channel open breast array coil (Invivo). MR performance was measured by scanning the phantom with both the Biograph mMR and a 3.0 T TIM Trio (Siemens Healthcare) using the following sequences: axial T1-weighted 3D (TR/TE=4.57/1.81 ms, flip angle=8.0°, TA=173 s), axial T2-weighted 2D (TR/TE=5310/99 ms, flip angle=90°, TA=258 s), and a 2D axial STIR (TR/TE=5390/99 ms, TI=220 ms, flip angle=80°, TA=244 s). Image uniformity was compared qualitatively and SNR was calculated by drawing circular ROIs (diameter=22 mm) on the phantom and background regions and taking the ratio of the mean and standard deviations computed from these ROIs, respectively. PET image quality was assessed by scanning the cylinder phantom filled with $^{18}$F for 120 minutes both with (start activity= 1.34 mCi) and without (start activity=1.37 mCi) the coil placed in the system. Images were reconstructed using the manufacture's implemented OP-OSEM (iterations=3, 4.0 mm FWHM Gaussian blurring in post-processing) with corrections for attenuation, scatter, randoms, dead-time, and decay applied. As the breast coil was not developed to be compatible with the PET component of the Biograph mMR we generated the image volume containing linear attenuation coefficients (LACs) at 511 keV of the coil (μ-map) specifically for this analysis [2]. The coil was scanned on the CT component of a Discovery STE PET/CT system (GE) (tube current=540 mA, tube voltage= 540 kVp), segmented to remove the patient table, and converted from HU to LACs at 511 keV using the bilinear transform method [3] (transition point=50 HU, lower threshold=-500 HU) to generate the μ-map. Spatial transformation of the coil μ-map into the Biograph mMR PET frame of reference was performed by scanning the breast coil "painted" with ~1 mCi of $^{18}$F for 120 minutes, reconstructing the emission data, and manually registering the μ-map to the "painted" coil images using an affine transform. Quantification of PET image uniformity was measured by drawing circular ROIs (diameter=10.8 cm) on a range of coronal slices covering the long axis of the cylinder and computing the bias with respect to the volume mean.

**Results and Discussion:** Image SNR for the MR component of the Biograph mMR and the 3.0 T TIM Trio was found to be comparable (Table I), and no significant artifacts were visible in Biograph mMR images. For PET performance, images with the coil in the FOV showed significantly improved uniformity with inclusion of the coil μ-map, however, quantification was reduced compared to the cylinder phantom scanned in the absence of the coil (Figures 1 and 2). For example, RMSE calculated over all slices was 3.9%, 7.7% and 2.2%, when the coil μ-map was either included or not, or when the coil was not placed in the FOV, respectively. Residual error between PET images, with or without the coil present in the FOV, was most significant at larger slice numbers corresponding to the largest quantity of coil material between the phantom and PET detectors. Artifacts due to beam hardening in the original CT images, spatial registration error of the coil μ-map into the PET frame of reference, and inherent inaccuracies in applying the bilinear transform to materials with significantly different properties then bone or soft tissue, may account for uniformity bias after all corrections.

**Conclusions:** We have performed a preliminary study to optimize image quality for breast imaging for both the MR and PET components of the Biograph mMR. Initial results suggest that MR performance of the Biograph mMR is comparable to stand alone systems. PET image uniformity for a coil not designed to be PET compatible (i.e. constructed to minimize attenuation of 511 keV photons) was significantly improved via the use of a calculated μ-map based on a CT acquisition, however, residual error remained even after all corrections. A PET compatible breast coil (similar to the 32 brain array recently developed (C.Y. Sander et al., *ISMRM annual meeting*, 2011)) would provide the most significant impact on image uniformity, however, the accuracy of the conventional coil μ-map may be improved via use of a industrial CT and improved segmentation methods. Future studies will be employed to optimize image quality of the coil during patient imaging.

**References:**

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**Table I.** MR SNR Comparison

| Sequence | SNR Biograph mMR | Trio |
|---|---|---|
| T1 | 257.7 | 243.7 |
| T2 | 55.6 | 71 |
| STIR | 58.3 | 49.3 |

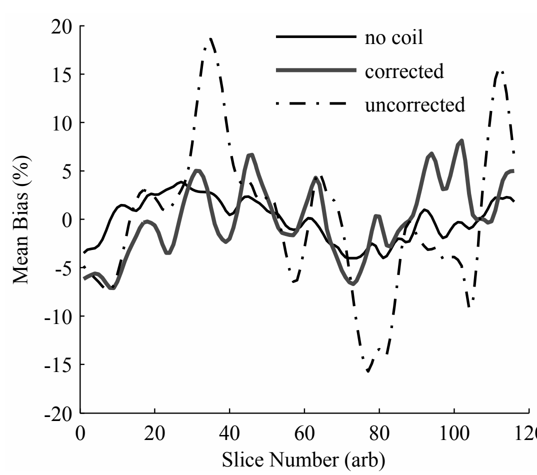


Figure 1. PET image uniformity along the long axis of the phantom.

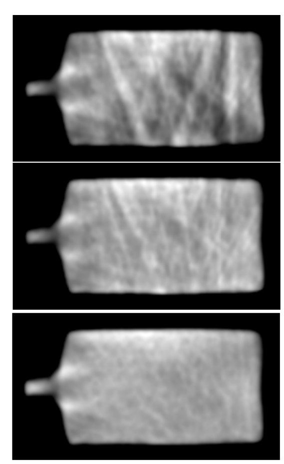

Figure 2. Axial PET images of the uniform phantom (from top) with the coil in the FOV and without or with the coil μ-map, and without the coil in the FOV.