Introduction: Positron emission tomography (PET) requires photon attenuation correction to accurately reconstruct the PET images. PET-MR systems use MRI to image and differentiate tissue and generate attenuation or $\mu$ (mu-) maps. Tissues are typically delineated into four categories (air, bone, fat, and water) using Dixon fat/water or ultra-short TE (UTE) pulse sequences. Unfortunately, these sequences have limitations in differentiating tissue types and foreign objects. Mu values can be accurately predicted from acquired and published tissue properties with the exception of solid bone which cannot be directly imaged with MRI due to its short T$_1$ (< 20 μs). The study’s goal was to identify the relationship between tissue properties and mu to optimize the accuracy of the mu-map.

Methods: T$_1$-weighted MPRAGE (TI: 0.9 s, TE:3 ms, TR:2.3 s), spin echo (TE:8.4 ms, TR:0.7 s), and UTE (TE:0.07/2.5 ms, TR:11.9 ms) MRIs; and proton density/T$_2$ (TEs:7.6/91 ms, TR:16 s) weighted MRI signals were acquired in human subjects (N=5, <age>= 80 years) on a Siemens 3T mMR (PET/MR) scanner after informed consent. Relaxation times, fat/water compositions, proton and mass densities, and magnetic susceptibilities of tissues were obtained from the literature (Table 1). Mass attenuation coefficients were calculated from http://www.nist.gov/pml/data/xraycoef/index.cfm/. Tissue regions of interest (ROIs) were analyzed for signal in which cannot be directly measured using MRI. Mu can be calculated based on magnetic susceptibility alone ($R^2=0.967$). Adding T$_1$- or T$_2$-weighted data did not affect the outcome although combining 11 tissue parameters resulted in accurate calculation of mu for all ten tissue types ($R^2=1.000$). Obviously, many of the tissue parameters are correlated (e.g., fat/water fraction, mass and proton densities, $T_2$ signal intensity and relaxation times). Most of the measured proton densities were significantly lower than the literature. Diploë (marrow), bone, skin, and sinus measurements were vulnerable to partial volume effects during the ROI analysis.

Results and Discussion: Based on the regression, mu can be calculated for the ten tissue types based on mass density ($R^2=0.996$) which cannot be directly measured using MRI. Mu can be calculated based on magnetic susceptibility alone ($R^2=0.927$) or combined with proton density ($R^2=0.967$). Adding T$_1$- or T$_2$-weighted data did not affect the outcome although combining 11 tissue parameters resulted in accurate calculation of mu for all ten tissue types ($R^2=1.000$). Obviously, many of the tissue parameters are correlated (e.g., fat/water fraction, mass and proton densities, $T_2$ signal intensity and relaxation times). Most of the measured proton densities were significantly lower than the literature. Diploë (marrow), bone, skin, and sinus measurements were vulnerable to partial volume effects during the ROI analysis.

Bone can be differentiated from air through their magnetic susceptibilities since air/tissue boundaries will have significant magnetic inhomogeneities while bone/tissue interface will not. Phase or susceptibility maps can be combined with image segmentation to accurately identify bone and foreign bodies (e.g., implants) from air for improved mu-maps.