SNR Comparison of RF Coil Size for Ischemic Skin Imaging

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Introduction

Peripheral vascular disease often presents as non-healing skin lesions. There is a significant clinical need for early diagnosis to institute preventative therapy. While early morphologic changes can be seen on biopsy, there is a general reluctance to perform these on patients suspected of having ischemic tissue due to wound healing concerns. Physiologic evaluation is complicated by the complexities that govern skin perfusion for skin thermoregulation. Laser Doppler is the method currently employed for the bulk of clinical perfusion evaluation in the skin. However, this method must be used with care, since the method is depth-dependent, and local skin-temperature changes can significantly affect measurements [1].

Magnetic resonance imaging has strong potential for skin perfusion imaging. *Song et al.* have demonstrated high-resolution images [2,3]. While good for morphology, imaging time prevents evaluation of skin perfusion physiology. An ideal MR skin perfusion imaging system will need to combine high-resolution, high-speed imaging sequences with optimized receiver coils. Since fast imaging is needed, coils could also be cooled with liquid nitrogen to further improve SNR [4]. Since normal capillary fill time is on the order of 3-4 seconds, a temporal resolution of 3 seconds or less will allow visualization of skin perfusion dynamics. In this abstract, we focus on the selection of RF coil parameters for this application. Specifically, we seek to determine the optimal coil size by trading off between SNR gains and increasing significance of resistive losses as coils get smaller.

Methods

Since the vessels of the skin lie in the hypodermis layer, we built and tested three coils for imaging 4.6 mm to about 1 cm. We required a sensitivity depth of 1.5 cm, or minimum diameter of 0.6 inches. It is well known that a reduction in coil diameter leads to an increase in SNR of a factor of $d^{(-5/2)}$, where d is the coil diameter [5]. However, as coils get smaller and smaller, coil resistive losses become a more significant noise term, and eventually will contribute as much as the conductive tissue, or inductive, losses. We therefore must find the optimum size in this tradeoff for hypodermic skin imaging.

We constructed 3 coils of diameters 1, 1.25, and 1.5 inches, from 125 μm thick copper foil to ensure thickness above the skin depth. Each coil was tuned for 50 ohms at 63.86 MHz resonance. The loaded and unloaded Q's were measured, and their ratios were compared with the ratio of unloaded vs. loaded input resistance. At 1 and 1.25 inches, coil losses became a significant loss term compared with inductive losses. Not taking into account these coil losses, we made predictions for SNR improvement based on coil size alone; these predictions are listed in Table 1.

We then performed SNR measurements on the same volunteer, with each coil aligned to the same center location. Coils were placed along the quadriceps muscle so that the readout direction was perpendicular to the skin-coil interface. Coils were spaced from the skin by 3 mm using cardboard to minimize dielectric losses from capacitor fringing electric fields. Using a rapid 3D gradient-echo sequence, we acquired a 4×4×3-cm slab volume centered about the coil, with resolution of 156×208 microns in-plane and 0.7 mm in the slab direction. TE and TR were 5.6 and 28.8 ms, respectively, for a volume imaging time of 4:26. Once slices containing the same morphological markers were identified, regions of 24 pixels were chosen in the same location of the hypodermic fat for SNR measurements. The SNR was measured using the mean of these regions and the mean of 24 background pixels. To demonstrate sub-100 micron resolution, we performed rapid 3D gradient echo at the same location with a slab volume of 4×4×1.1 cm, with resolution of 78×78 μm × 1.0 mm, TE/TR of 9.6/55.9 ms, and a scan time of 7:38.

Results and Discussion

Table 1 shows the measured coil bench parameters corresponding to the relative resistive and inductive loss terms. SNR measurements from the hypodermic fat in the 156×208 μm × 0.7 mm sequence are also shown along with the SNR gain factors predicted by coil size. Figure 1 contains a 1 mm thick slice from the longer 3D rapid gradient echo sequence. Figure 2 shows an image from the shorter 3D rapid gradient echo volume. This is one slice of the acquired volume used for SNR measurement. Both of the images in Figures 1 and 2 were acquired using the 1-inch coil. Note that the measured relative SNR was increasingly lower than predicted SNR increase (by size alone) as the receive coil got smaller.

Conclusion

Although coil resistive losses increase significantly at the lowest coil size, the gain in SNR from having a smaller noise volume is still larger, resulting in a better receive coil SNR for imaging of the hypodermic layer of the skin for the coil sizes tested. We present images using the 1-inch coil with good delineation of the hypodermis and vessels, as well as the papillae in the dermis layer. We predict that SNR will be improved even more by cooling this coil with liquid nitrogen. The additional SNR gain coupled with a fast imaging sequence could lead to preliminary perfusion measurements in the skin for comparison with other modalities such as laser Doppler.

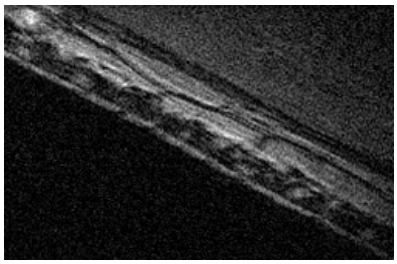

Figure 1. 1.5 x 2 cm image from 78 x 78 μm data, 1 mm slice near the quadriceps of a normal volunteer.

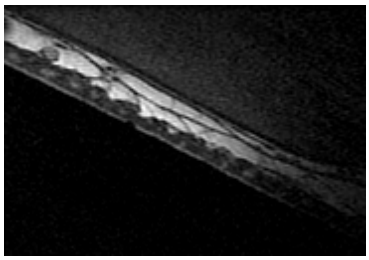

Figure 2. 2.5 x 3 cm image from 156 x 208 μm data with 0.7 mm slice thickness from the 3D data set used for SNR measurements near the quadriceps of a normal volunteer.

Table 1: Measured and predicted parameters for coil comparison

Coil diameter, inches	1.5	1.25	1
Unloaded Q	213	164	193
Loaded Q	118	120	152
Relative SNR	1	1.22	1.72
Predicted SNR (based on coil size alone)	1	1.58	2.76

References

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