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Magnetic field shimming of a permanent magnet using a combination of pieces of permanent magnets and a single-channel shim coil for skeletal age assessment of children

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ABSTRACT

We adopted a combination of pieces of permanent magnets and a single-channel (SC) shim coil to shim the magnetic field in a magnetic resonance imaging system dedicated for skeletal age assessment of children. The target magnet was a 0.3-T open and compact permanent magnet tailored to the hand imaging of young children. The homogeneity of the magnetic field was first improved by shimming using pieces of permanent magnets. The residual local inhomogeneity was then compensated for by shimming using the SC shim coil. The effectiveness of the shimming was measured by imaging the left hands of human subjects and evaluating the image quality. The magnetic resonance images for the child subject clearly visualized anatomical structures of all bones necessary for skeletal age assessment, demonstrating the usefulness of combined shimming.

1. Introduction

Skeletal age, which gives a measure of child growth, is evaluated by assessing the maturity of bones in the left hand and wrist [1–3]. Although plain radiography has been the gold standard for skeletal age assessment, magnetic resonance imaging (MRI) has recently emerged as an alternative [4–6] because of its noninvasive and nonirradiative nature. In a previous study [7], we showed the validity of skeletal age assessment using an open, compact MRI system. However, the magnetic circuit used in the previous work was designed to image an adult hand [8–10], and was large for young children (620 mm × 1006 mm × 620 mm in size, 700 kg in weight). Indeed, in some examinations, the arm of the subject was too short to reach the center of the imaging volume, and the unnatural posture of the body introduced a severe motion artifact that hampered the skeletal age rating. To remedy this problem, in this study, we used a smaller permanent magnet and developed a new, dedicated MRI system for the skeletal age assessment of children.

From a technical point of view, high homogeneity of the magnetic field in the imaging volume is required to obtain high-quality magnetic resonance (MR) images, because inhomogeneity leads to image artifacts such as signal loss and image distortion. The field homogeneity of a permanent magnetic circuit immediately after installation, however, is generally far from the imaging criteria, because of the manufacturing tolerance and variations and the influence of surrounding objects. Therefore, magnetic field shimming is required to compensate for the field inhomogeneity to meet the target imaging criteria.

The standard approach for passively shimming permanent magnets uses movable blocks or additional small pieces of permanent magnets. High homogeneity can be achieved by precisely positioning a number of shim pieces of permanent magnets. Different methodologies of passive shimming are being developed by a number of research and development groups, and there have been remarkable improvements in field homogeneity, especially in the fields of portable NMR/MRI with single-sided [11–16] and closed [17–22] permanent magnets. The details including instrumentations and applications are reviewed elsewhere [23–26]. For example, the use of shim units built from movable permanent blocks and fine control of their positioning allow sub-ppm spectroscopic resolution sufficient to measure ^{1}H NMR spectra, in a single-sided magnet [13] and in a Halbach closed magnet [20]. In contrast, relatively few reports concerning passive shimming for open-type biplanar permanent magnets have been published. For such magnets, small pieces of magnetic material (iron/permanent magnets) are attached to a pole face for the fine-tuning of field homogeneity with the use of an iterative optimization algorithm [27–30].

The field homogeneity that can be obtained with passive shimming is limited by undesired errors in positioning shim magnet pieces. The higher homogeneity can be obtained by additional use of electrical shim coils [31], which further reduce residual field inhomogeneity beyond the limit of passive shimming. Furthermore, the homogeneity limit of passive and electrical shimming

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(or hardware shimming) can further be corrected by using sophisticated NMR measurement schemes such as nutation echoes [32] and shim pulse techniques [33–35].

Electrical current shims use multiple coils that are individually designed to correct a specific spherical harmonic distribution in field inhomogeneity. In some ideal cases, the spatial distribution of the magnetic field is stationary, and amplitudes of currents in shim coils are fixed. Then, multiple shim coils could be simplified to a single-channel (SC) shim coil that has a pattern designed by calculating the superposition of currents in the multiple shim coils that are determined to compensate for a given inhomogeneity. The concept of SC shimming has been used in the design of a biplanar shim coil for Halbach magnets [36–38].

SC shimming has the advantages of simplicity of hardware and installation space and could be an alternative to multichannel electrical shimming. Furthermore, the design and fabrication of SC shim coils are relatively simple and easy, whereas fine shimming with small pieces of permanent magnets requires sophisticated techniques and much effort to maintain accuracy of positioning during the iterative process of shimming.

Like conventional electrical shimming, however, SC shimming has an upper limit of the correctable field strength and is not suitable for the compensation of large field inhomogeneities. This is because the current in an SC coil should be limited so as not to cause excess heat generation that degrades the field homogeneity. Therefore, it is practical and desirable to make use of SC shimming combined with coarse passive shimming with permanent magnet pieces.

In this study, the combined shimming was tested with experimental measurements of the magnetic field and evaluation of the quality of MR images. We first performed coarse shimming with pieces of small permanent magnets and then performed SC shimming to correct the residual high-order spatial inhomogeneity. In addition, we obtain MR images of the left hands of human subjects with the sequences used for MR skeletal examination in the previous study [7] and show that image artifacts are sufficiently suppressed so that anatomical structures of bones necessary for skeletal age assessment are clearly visualized in the MR images.

2. Experimental

2.1. Compact MRI system

Fig. 1a and b are an overview and a photograph of the compact MRI system except for the MRI console. The spatial coordinates were defined as shown in the figure. The compact MRI system consisted of a C-type Nd–Fe–B permanent magnet (Shin-Etsu, Chemical Co. Ltd., Tokyo, Japan), a gradient coil set, an RF probe, and an MRI console. The specifications of the permanent magnet were: a field strength of 0.3 T, gap width of 120 mm, dimensions of 400 mm × 570 mm × 410 mm, and weight of 450 kg. This magnetic circuit was roughly shimmed using Nd–Fe–B magnet plates by the manufacturer, and the magnetic field homogeneity was not high (350 ppm (root mean square, RMS) and 1500 ppm (peak-to-peak, PP) over 160 mm × 160 mm × 50 mm diameter ellipsoidal volume (DEV)). The magnet was kept at 28 °C because the magnetic field strength and spatial variation were sensitive to temperature changes [38].

We used a solenoid RF coil optimized for imaging a child's hand that was the same as the coil used in the previous work [7]. The coil was a 16-turn solenoid, 17.6 cm long, and was made by winding Cu foil (0.1 mm thick) around an oval acrylic pipe (aperture: 10 cm × 5 cm; length of 220 mm; thickness of 4 mm) and divided in quarters with three chip capacitors (100 pF) to reduce stray capacitance between the hand and coil. The RF coil was shielded by a rectangular RF probe box made of 0.3-mm-thick brass plates. A 5-mm-thick aluminum plate was connected to the outside of the brass box to ground the arm and thus minimize interference by external RF noise [8]. The typical duration of the 90° hard pulse was 120 μs using an output power of 200 W.

2.2. Gradient coil design

The *x*- and *y*-gradient coils were designed as a combination of a circular arc and third-order Bezier curve with the position and center angle optimized using a genetic algorithm with a minimal generation gap model (GA/MGG) [39,40]. The maximum number of turns was 30 and the coil gap was set to 120 mm. The *z*-gradient coil was designed as a combination of circular current loops with diameter optimized using a GA/MGG. The maximum number of turns was 40 and the coil gap was set to 105 mm. In each calculation, the wire diameter was set to 1 mm and the coil pattern was restricted to a circular region with a diameter of 320 mm. The number of iterations was 100,000 and the total calculation time was almost a week for the *x*- and *y*-gradient coils and three days for the *z*-gradient coil using a 2.7-GHz clock-frequency Pentium dual-core processor. The calculated nonlinearity of the gradient field of the *x*-, *y*-, and *z*-gradient coils was 9.1%, 9.1%, and 7.6% in 180 mm × 180 mm × 60 mm DEV, respectively.

Each gradient coil element was made by winding polyethylene-coated Cu wire (1 mm in diameter) on the surface of a fiber-reinforced plastic (FRP) plate (360 mm in diameter, 0.5 mm thick) and tracing a printed coil pattern attached on the other side of the plate (Fig. 1c); *x*, *y*, and *z* coil elements were then stacked together using epoxy resin. The gradient coil elements were driven by a three-channel gradient driver (±10 A, DGD10AT, Digital Signal Technology, Inc., Saitama, Japan). The measured efficiency of the *x*-, *y*-, and *z*-gradient coils was 2.28, 2.32, and 3.11 mT/m/A, the resistance was 1.4, 1.4, and 1.2 Ω, and the inductance was 260, 250, and 390 μH, respectively.

2.3. Target region of interest for shimming

The typical size necessary for the MR skeletal examination of children aged 12 years is 80 mm (x) × 155 mm (y) × 25 mm (z). Accordingly, we determined the target region of interest (ROI) for shimming to be a 120 mm × 160 mm × 50 mm DEV.

2.4. Evaluation of the magnetic field inhomogeneity

The spatial variation in the magnetic field B_0 was measured using two methods according to the order of the homogeneity. When the B_0 homogeneity was low at the initial stages of shimming, the variation was measured by a nuclear magnetic resonance (NMR) probe. The probe consisted of a solenoid RF coil wound around a 6-mm diameter glass sphere filled with baby oil (Johnson & Johnson, Skillman, NJ, USA), tuning and matching capacitors, and a rectangular shield box made of brass plates (50 mm × 60 mm × 25 mm). To measure the B_0 inhomogeneity, the NMR probe was positioned in the magnet gap by a three-axis stage with accuracy of 0.1 mm, and the free induction decay (FID) signal was measured at the given position. The magnetic field was then determined by the peak center of the Fourier transform of the FID signal. Likewise, the magnetic fields at grid points 1 cm apart in a rectangular area (160 mm × 160 mm) were measured.

When the homogeneity was improved to a high level, the B_0 spatial variation was evaluated employing a conventional MRI-based, phase-shift method with a phantom made of a cylindrical acrylic container (200 mm in diameter, 55 mm high) filled with $CuSO_4$-doped water. This method is more accurate and efficient than the above NMR probe-based method.