Theoretical and Experimental Evaluation of Detached Endcaps for 3 T Birdcage Coils

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The use of detached endcaps for 3 T birdcage coils was investigated both theoretically and experimentally. Finite difference time domain analysis, along with workbench and MRI techniques, were used to map the radiofrequency (RF) B1 distribution along the coil axis with and without an endcap. Without an endcap the measured B_1 value at the service end of the birdcage was only 45% of the value at the coil's center. This was improved to 85% with a detached endcap of maximum achievable diameter (375 mm), positioned 4 mm from the RF shield. The B_1 field distribution on the patient side of the coil was unaffected by the presence of the endcap. The dependence of the B_1 distribution as a function of endcap diameter was also investigated. Surprisingly, simulations and experiments show that there is an optimum ratio of endcap-to-birdcage coil diameter (\approx 1.08) that gives the best B_1 homogeneity. In the human head the optimized endcap, positioned 16 mm from the RF shield, improves the MRI signal amplitude from 55% to 85% of maximum toward the service end. This novel endcap design is easy to implement with existing birdcage coils, and could prove useful when flexibility in access to the RF coil is required. Magn Reson Med 49:363-370, 2003. © 2003 Wiley-Liss, Inc.

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Birdcage coils (1) with a length twice their diameter provide a reasonably homogeneous RF (B_1) field at low static magnetic field (≤ 1.5 T), and as such are widely used for clinical MRI studies. However, anatomical constraints of ten require that a short birdcage coil (total length < 35 cm) be used for human brain studies, which can lead to additional intrinsic B_1 inhomogeneities along the coil *z*-axis and in the transverse plane. This can be further compounded at higher field strengths where the effective RF wavelength in the human head is comparable to or shorter than the length of the coil elements (typically < 25 cm). Under such conditions, wavelength effects in the RF currents along the coil elements lead to further degradation of the RF homogeneity, and hence to a significant B_1 inhomogeneity along the coil axis (sagittal/coronal plane) (2).

Recently, we reported studies of the RF B_1 distribution in cylindrical phantoms of varying electrical properties, and in head studies of healthy volunteers using a 3 T MRI research system equipped with a birdcage coil (3). Theory and experiments showed that along the birdcage coil zaxis the B_1 distribution is determined mainly by the RF coil geometry. These results suggest that hardware methods for improving the intrinsic RF homogeneity along the z-axis of head-size birdcage coils should be investigated at 3 T and above. For example, it was recently shown that active transmit power modulation as a function of axial slice position is effective in compensating for B_1 field inhomogeneity at 3 T (4). In this study we investigated an alternative hardware optimization approach, using the RF coil endcap.

BACKGROUND

At low static field strength, the use of a conducting endcap attached to the service end of head-size RF slotted-cylinder (5) and birdcage (6,7) coils improves the RF homogeneity (note that we define the service end to be the direction farthest from the subject, which is generally the end of the coil used to attach the coaxial cabling). The endcap design is based on the theory of "current images" (8,9). This states that 1) the effect of an infinite ground plane for a filamentary conductor +i perpendicular to the ground plane is equivalent to having a second perpendicular current +i positioned below the ground plane; and 2) a conductor carrying a current +i parallel to, and at a height habove, a metal ground plane is equivalent to having a current -i, at a height *h*, below the ground plane. In accordance with the current-images method, a coil endcap acts like an RF mirror, and since the RF currents in the legs of the birdcage are perpendicular to the endcap, the effective electrical length of the coil is doubled. However, with the presence of an endcap there is also a "spurious" RF field contribution generated by the image of the end-ring RF currents. Thus, a full characterization of the RF homogeneity of a birdcage coil toward its service end requires that the combination of these two factors be considered. Additionally, the method of current images can only provide an accurate estimate of mirror currents for the case of a ground plane of infinite extent.

Typically, RF coil endcaps are constructed with a copper foil disk that is an integral part of the service end of the birdcage coil (permanent endcap). According to conventional thinking, in order to optimize the B_1 distribution along the coil (z) axis the diameter of the endcap (D_e) should be very large with respect to the birdcage coil diameter (D_c). However, some functional MRI (fMRI) studies may require optimized gradient performance for which a small-diameter (typical inside diameter < 40 cm) headinsert gradient coil can be used. This constrains the endcap diameter to be comparable to the birdcage dimensions,

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and standard analytical models (e.g., application of the Biot-Savart law and the current-image method) for calculating the B_1 distribution in the presence of the endcap are inaccurate. The inaccuracy increases at high field (≥ 3 T) where the effective RF wavelength in the presence of the human head is comparable to or smaller than the coil size. For these reasons, at high field a 3D full-wave solution of Maxwell's equations is required to calculate the RF field distribution of a birdcage coil with and without an endcap. Recently, a finite difference time domain (FDTD) analysis of endcapped birdcage coils was performed at 1.5 T (10–12). However, in those studies only permanent endcaps were considered and/or the RF shield was not included in the FDTD model of the coil.

The use of an endcap positioned at a given distance from the service end of the birdcage coil (detached endcap) has been suggested as an alternative method to improve the RF homogeneity (13). This design is particularly interesting for applications in which full access to the coil's service end may occasionally be required for certain experimental conditions. However, only a simplified Biot-Savart analysis was described previously (13); the end-rings and RF shield were not included in the model, and no experimental data were reported. We recently reported a preliminary extended study of detached endcaps for 3 T birdcage coils (14–16).

In the present study we investigated theoretically and experimentally the design of detached endcaps for 3 T birdcage coils. We propose a new endcap design that allows optimization of the RF B_1 field distribution along the axis of birdcage coils. This is achieved by adjusting the diameter of the endcap for a given distance between the endcap and the birdcage coil. An FDTD analysis, along with workbench and MRI techniques, were employed to test this novel endcap design.

MATERIALS AND METHODS

3 T Scanner

The 3 T human MRI research system is composed of a 1-m-bore superconducting magnet (Oxford Magnet Technology, Oxford, UK) connected to a Varian Unity Inova console. A Magnex SGRAD MKIII head gradient coil (Magnex, Ltd, Oxfordshire, UK) is housed within the bore, and is driven by Siemens GPS amplifiers (600V, 250A). The gradient coil has an inside diameter of 38 cm, and provides a maximum gradient strength of 34 mT/m in a minimum of 200 μ s. This system includes a 4 kW RF amplifier (model 4T40; American Microwave Technology).

RF Birdcage Coil

Our 3 T system is equipped with a head-size, 16-element, quadrature birdcage coil, used for both pulse transmission and signal reception. This birdcage coil, based on a hybrid design (17), is tuned to 127.3 MHz and was used for workbench and MRI testing. This birdcage coil has a diameter of 278 mm, a length of 210 mm, and an RF copper shield of dimensions 376 mm diameter, 260 mm length, and 5 μ m thickness. The RF shield diameter could not be made larger because of the restricted access of the head gradient coil. The center of the birdcage coil RF shield was



FIG. 1. **a:** Schematic diagram of the gradient coil, birdcage coil, RF shield, and detached endcap. **b:** Photograph of the slotted endcap prototype ($D_e = 304 \text{ mm}$, d = 12 mm) when attached to the service end of the experimental birdcage coil.

shifted 16 mm toward the patient end of the coil, i.e., the service end of the birdcage is at 9 mm from the end of the RF shield. The minimum attainable distance, d_0 , between the end of the RF shield (z = 0) and the endcap was 4 mm.

Endcap Design

A schematic diagram of the gradient coil, birdcage coil, RF shield, and detachable endcap is shown in Fig. 1a, where $D_{\rm e}$ is the endcap diameter and d is the distance between the endcap and the service end of the RF shield (z = 0). We have built and tested several detachable endcap prototypes suitable for workbench and MRI calibration. Two small holes (diameter 2 cm) were made in the endcaps to allow for quadrature driving of the birdcage coil at its service end. Small plastic screws were used to position the endcaps at the service end of the birdcage coil.

First, a detachable endcap made of copper foil ($D_e = 375 \text{ mm}$, thickness 35 µm) fixed on an acrylic disk (diameter 375 mm) was built and used for workbench and MRI calibration. This endcap was used to study the dependence of the B_1 distribution as a function of the distance d from the service end. Values of d from 4 mm to 34 mm were investigated. We have also studied the dependence of the B_1 distribution as a function of D_e . To this purpose, additional endcap prototypes were constructed as above with values of $D_e = 80$, 152, 200, 232, 264, 282, 298, 311, 327, 344, 360, and 375 mm. Larger values of D_e could not

Table 1 Birdcage Coil RF Parameters When Empty as a Function of Endcap Position *d* for the Slotted Endcap

<i>d</i> (mm)	f _o (MHz)	Qª	S ₁₁ (dB)	Isolation (dB)
6	128.873	50	-9	-11
11	128.348	50	-10	-12
18	127.983	50	-11	-14
25	127.707	50	-12	-15
No endcap	127.373	60	-15	-17

*The endcap diameter was fixed at D = 304 mm.

^aMeasured from the reflection coefficient (S₁₁) response as twice the ratio of the resonant frequency to the frequency bandwidth at -3 dB. The Q values were not corrected for the relatively poor matching.

be tested in the magnet due to the limited access of the head-size gradient coil.

Significant eddy currents can be induced in the endcap by the fast-switching gradient coils. Several strategies are known to reduce or eliminate eddy currents induced in RF coils (18). The simplest design is an endcap made of ultrathin continuous copper foil. However, even if the copper thickness is limited to five skin depths (about 30 µm at 128 MHz), our experience shows that significant eddy currents are induced in the endcap. In this work, an optimized detachable slotted endcap was also built using printed circuit board (PCB) technology. This endcap (external diameter 304 mm, internal diameter 25 mm) was made of copper (thickness 35 µm) on a rigid fiberglass substrate (thickness 1.6 mm). To eliminate eddy currents, 16 slots were cut radially (width 3 mm). To maintain electrical continuity at 127.3 MHz, a number of chip capacitors (total capacitance of 900 pF) were connected across each slot. Figure 1b shows a picture of the slotted endcap prototype when attached to the birdcage coil. Some care is necessary in the design of slotted endcaps at 3 T. In fact, we found that slotted endcaps can act like a RF surface coil, giving rise to a number of "spurious" resonant modes. Depending on the specific endcap design, the presence of a spurious mode close to 127.3 MHz can produce a frequency displacement of the birdcage coil. This can be explained as a mutual resonant inductive coupling mechanism (19). Experimentally we found that the perturbation of the birdcage coil was avoided by increasing to 2900 pF the capacitance across two diagonally opposite slots. It is worth noting that similar effects are expected if the endcap is made of tiled rectangular copper foil elements separated by a thin dielectric.

This slotted endcap prototype was attached to an acrylic disk (diameter 375 mm), fitted to the birdcage coil and tested with phantoms and volunteers using MRI pulse sequences. The network analyzer was used to measure the reflection coefficient (S_{11}) , the resonant frequency (f_0) , the quality factor (Q), and the quadrature isolation (I) of the birdcage prototype (when positioned within the scanner) with and without the slotted endcap. The experimental results obtained for *d*-values of 6–25 mm are reported in Table 1.

3D FDTD Simulation

At 3 T a realistic simulation of the RF B_1 distribution in the birdcage coil with and without the endcap requires pow-

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erful EM computational techniques. Several numerical EM techniques have been developed to evaluate the B_1 distribution in birdcage coils (13). In the present study, the FDTD method was used to calculate the 3D B_1 field distribution of the empty birdcage with and without the endcap. This method was described in detail previously (20). Steady-state solutions of the transverse $(x,y) B_1$ field magnitude were calculated at 128 MHz with the FDTD method at regular intervals along the coil axis. The Yee cell dimensions were 4-mm resolution in the x-oriented (anterior-posterior) and y-oriented (left-right) directions, and 2.5-mm resolution in the z-oriented (inferior-superior) direction. All calculations were set up and performed with the aid of commercially available software ("xfdtd"; Remcom, Inc., State College, PA).

The simulated data were obtained by considering an ideal low-pass 16-element birdcage coil, RF shield, and endcap with dimensions and relative locations similar to those of the experimental coil. The model birdcage coil had a diameter of 278 mm and a length of 221 mm. An RF copper shield was also modeled (diameter 380 mm, length 254 mm, thickness 1 Yee cell). The center of the RF shield was shifted by 16.5 mm toward the patient end, relative to the center of the birdcage coil elements, i.e., the service end of the birdcage coincides with the service end of the RF shield (z = 0). This FDTD model matches more closely a previous experimental birdcage prototype used for accurate workbench calibrations (14–16).

First, the FDTD simulation was run for the case of an endcap of constant diameter $D_e = 380$ mm positioned at a distance d of 2.5–22.5 mm. Other intermediate distances, including the case for no endcap, were also modeled. Second, the FDTD method was used to map the B_1 distribution for the birdcage coil without the endcap and with endcaps of diameter D_{e} of 76–780 mm. In these latter FDTD simulations the endcaps were maintained at d =2.5, 12.5, and 22.5 mm in turn. Third, the FDTD technique was used to calculate the B_1 distribution for the unshielded birdcage coil without the endcap and with endcaps of diameter D_e of 76–780 mm (with d = 12.5 mm). To estimate the improvement due to the endcaps, a B_1 gain parameter was calculated as $G = [(B_1^*(z)/B_1(z))-1]$, where $B_1^*(z)$ and $B_1(z)$ are the B_1 amplitude at a given axial position with and without the endcap, respectively.

MRI Testing With Phantoms

To experimentally evaluate the RF B_1 field distribution of the birdcage coil prototype with and without the endcaps, we used a plastic bottle (diameter 17 cm, length 26 cm) filled with oil (Esso Bayoil, Esso, UK). This low dielectric constant ($\epsilon_r = 2.9$) phantom does not significantly perturb the *intrinsic* RF B_1 field distribution, nor the Q and impedance matching of the birdcage coil (3). To assess B_1 homogeneity fast low-angle shot (FLASH) gradient-echo images were collected in the sagittal, coronal, and axial orientations with and without the experimental endcaps. The acquisition parameters were: TE = 6 ms, TR = 60 ms, flip angle = 15° , FOV = $35 \text{ cm} \times 35 \text{ cm}$, slice thickness = 3 mm, and 256×256 pixels. Because of the low flip angle used the MRI signal amplitude is proportional, to first order, to the square of the B_1 field amplitude (a factor of B_1 due to the transmitted flip angle, and a factor of B_1 due to the receive sensitivity). The short TR (60 ms) introduces some minimal T_1 relaxation effects in the MRI images. However, since TR was held constant in all the experiments with the oil phantom, we are able to compare the B_1 field distribution with and without the endcap.

To study the dependence of the B_1 distribution as a function of the coil-to-endcap distance, MRI images were acquired with the copper endcap ($D_e = 375$ mm) positioned at d = 4, 10, 16, 22, 28, and 34 mm. Second, to study the B_1 distribution as a function of endcap diameter D_e , MRI images were acquired with the endcap at a fixed distance (d = 16 mm) and with $D_e = 80, 152, 200, 232, 264, 282, 298, 311, 327, 344, 360, and 375 mm. From these MRI images the <math>B_1$ field distributions along the dimension of the coil axis were obtained. We also obtained the gain parameter G from data acquired with and without the endcaps present.

In Vivo MRI Testing

FLASH gradient-echo images from volunteers were collected in the sagittal, coronal, and axial orientations, with and without the slotted endcap ($D_{\rm e} = 304$ mm, d = 12 mm). The acquisition parameters were: TE = 6 ms, TR = 60 ms, flip angle = 20°, FOV = 35 cm × 35 cm, slice thickness = 10 mm, and 256 × 256 pixels. From these images, MRI signal plots along the inferior–superior direction were obtained and used to estimate the improvement due to the endcap. Informed consent was obtained from all volunteers, and the project was approved by the local ethics committee.

RESULTS AND DISCUSSION

Workbench Calibration

From the measured S_{11} responses of the birdcage coil loaded with the oil phantom we found that: 1) without the endcap it was $f_0 = 126.95$ MHz, $S_{11} = -4$ dB, Q = 97 (not corrected for poor coupling), and I = -17 dB; and 2) with the endcap ($D_e = 375$ mm, d = 4 mm) the values were: $f_0 =$ 129.18 MHz, $S_{11} = -12$ dB, Q = 51, and I = -11 dB. Because of mutual inductive coupling, the presence of the endcap at the minimum value of d = 4 mm increased the resonant frequency of the birdcage coil by about 2.2 MHz. However, this frequency shift is within the tuning range of the birdcage coil. As shown in Table 1, similar variations of the RF parameters of the empty birdcage coil were also observed in the presence of the slotted endcap for a range of endcap distances.

B_1 Mapping vs. Endcap Distance

Figure 2a reports the numerically calculated FDTD B_1 distributions along the coil axis of the shielded model birdcage coil obtained with and without the endcap ($D_e = 380 \text{ mm}$) positioned between d = 2.5 and 22.5 mm. For comparison purposes the B_1 amplitude at the center of the coil was normalized to 100%. Without an endcap present the RF B_1 field showed a severe inhomogeneity, with the maximum B_1 at about z = -11 cm from the service end (service end defined to be z = 0 cm). At the service end



FIG. 2. **a:** Numerically calculated FDTD RF B_1 distributions along the *z*-direction for a model of the empty birdcage coil obtained without (continuous line) and with the endcap ($D_e = 380$ mm) positioned at d = 22.5 mm (circles), d = 12.5 mm (diamonds), d =5 mm (triangles), and d = 2.5 mm (squares). For comparison purposes the B_1 amplitude at the center of the coil was normalized to 100%. The coordinate z = 0 corresponds to the service end of the RF coil. **b:** Experimentally measured *z*-direction RF B_1 distribution for the birdcage coil obtained without (continuous line) and with the endcap ($D_e = 375$ mm) positioned at d = 28 mm (circles), d =16 mm (diamonds), d = 10 mm (triangles), and d = 4 mm (squares). The birdcage coil was loaded with the oil phantom, and for comparison purposes the B_1 amplitude at the center of the coil was normalized to 100%.

itself the B_1 value was calculated to be only about 50% of the maximum value. The numerically simulated results of Fig. 2a also show that with the detached endcap, a better B_1 amplitude is obtained as the endcap is positioned progressively closer to the service end. The B_1 amplitude increases up to about 98% of maximum at the service end with d = 2.5 mm. As expected, the B_1 distribution on the patient side of the coil was practically unaffected by the presence of the endcap.

Figure 2b shows the measured B_1 distributions along the coil axis obtained for the experimental birdcage coil with and without the endcap positioned between d = 4 and d = 34 mm. Again, the B_1 amplitude at the center of the coil was normalized to 100%. Without the endcap the RF B_1 field in the z dimension showed a severe inhomogeneity, with the maximum B_1 also at about -11 cm from the



FIG. 3. **a:** Numerically calculated FDTD RF B_1 amplitude at z = -0.5 cm (circles), z = -1.5 cm (continuous line), and z = -2.5 cm (squares) as a function of endcap position *d*. Data were obtained for the empty model birdcage coil and the endcap with $D_e = 380$ mm. **b:** Experimentally measured RF B_1 amplitude at z = -0.6 cm (circles), z = -1.5 cm (continuous line), and z = -2.5 cm (squares) as a function of endcap position *d*. Data were obtained for the birdcage coil loaded with the oil phantom and the endcap prototype with $D_e = 375$ mm.

service end. At the service end itself the B_1 value was only about 40% of the maximum value. In accordance with the FDTD simulations, the experimental results of Fig. 2b show an improved B_1 homogeneity along the z dimension as the detached endcap is positioned progressively closer to the service end of the coil. The B_1 amplitude increases up to about 85% of maximum at the service end with d =4 mm.

A full characterization of the numerically calculated FDTD B_1 amplitude for the empty model birdcage coil as a function of d is reported in Fig. 3a. As the endcap ($D_e = 380 \text{ mm}$) is moved toward the service end, the B_1 amplitude at the service end increases linearly, and for d = 0 the extrapolated B_1 amplitude is 100%. This extrapolated B_1 amplitude is in agreement with previous FDTD calculations of *permanent* endcap designs at 1.5 T (5–6, 9–11). The FDTD calculations suggest that a near-optimal B_1 distribution along z could be obtained with $d \leq 4$ mm. With the present experimental birdcage prototype, such short endcap-to-shield distances could not be tested. Note, however, that for smaller distances a larger detuning and mismatching of the birdcage coil would be expected, which

may restrict the practical usefulness of detached designs at such small values of d. The measured B_1 value in proximity to the service end of the experimental birdcage coil loaded with the oil phantom as a function of d is reported in Fig. 3b. The B_1 amplitude increases linearly, and the extrapolated B_1 amplitude for d = 0 is about 90%. This extrapolated value is smaller (about 10% difference) as compared to the FDTD simulated value.

Small discrepancies are observed between the simulated and experimental results of Figs. 2 and 3. However, this is not surprising considering the differences between the geometrical size of the model and the experimental birdcage coils, and the T_1 relaxation effects due to the short TR. A better agreement (max B_1 difference < 10%) was previously reported for workbench and FDTD data (15) when the more similar previous experimental birdcage coil was used.

B₁ Mapping vs. Endcap Diameter

The calculated FDTD RF B_1 distributions along z for the empty model birdcage coil obtained without the endcap and with the endcap of small, medium, and large diameter are reported in Fig. 4a. In these simulations the endcap was positioned at d = 12.5 mm, selected as a value that is anticipated to require only a modest retuning of experimental coils. As before, without the endcap the RF B_1 field along the z dimension shows the maximum B_1 at about -11 cm, and the B_1 amplitude at the service end was about 45% of the maximum. Surprisingly, the numerical results of Fig. 4a show that the intermediate diameter endcap $(D_{e} = 304 \text{ mm})$ gives a better B_{1} homogeneity along z than the largest-diameter endcap. In fact, for the intermediate diameter the B_1 amplitude increases to within 90% of maximum at the service end. The B_1 distribution on the patient side of the coil was practically unaffected by the presence of the endcaps. These numerically simulated results also show that an increase in the peak B_1 amplitude (about 10%) is expected with the intermediate endcap diameter. For comparison, the experimentally measured RF $B_{\rm 1}$ distribution along z for the birdcage coil prototype loaded with the oil phantom obtained without the endcap and with the endcap of small, medium, and large diameter are reported in Fig. 4b. The experimental MRI data were obtained with the endcaps positioned at d = 16 mm.

To estimate the optimum endcap diameter, the gain parameter G as a function of endcap diameter was calculated at several locations along the coil axis from the simulated and experimental data, and the results are reported in Fig. 5. We observe from Fig. 5a that at the service end (z = 0) a large endcap diameter ($D_e > 400 \text{ mm}$) gives a theoretical gain of about 44%. However, if the endcap diameter is reduced to $D_{\rm e} = 304$ mm, the gain *increases* to about 58%. For small diameters ($D_{\rm e} < 250$ mm) the gain decreases monotonically to zero. As shown in Fig. 5b, similar results were obtained for axial locations between the coil center (z = -11 cm) and the service end (z = 0) of the experimental birdcage coil. Endcap diameters larger than 375 mm could not be tested experimentally because of the limited access of the head-size gradient coil (free bore < 380 mm). The experimental results are in good agreement with the simulation, considering the differences in geometrical dimensions between the modeled and experimental birdcage coils, and the intrinsic errors in any experimental measurements. Figure 6 shows the simulated gain parameter *G* as a function of endcap diameter, calculated for several endcap distances (d = 2.5, 12.5, and22.5 mm). We observe that the maximum gain parameter value was obtained for the same endcap diameter of $D_{\rm e} \approx$ 300 mm.

Surprisingly, from these simulation and experimental results we conclude that there is an optimum ratio, $D_{\rm e}/D_{\rm c} \approx 1.08$, that gives the best B_1 homogeneity along the z-direction. The simulated FDTD results suggest that the optimum ratio $D_{\rm e}/D_{\rm c}$ is independent of the endcap distance, at least within the range evaluated in our study. It is worth noting that the gain improvement due to the optimized endcap (relative to a reference gain obtainable with the endcap of maximum $D_{\rm e} = 800$ mm) increases as the endcap distance is decreased. In fact, as shown in Fig. 6, the gain improvement is about 12% with d = 22.5 mm, 18% with d = 12.5 mm, and 25% with d = 2.5 mm.

From the FDTD simulations of the *unshielded* birdcage coil without the endcap and with endcaps of diameter $D_{\rm e}$



FIG. 4. **a**: Numerically calculated *z*-direction FDTD RF B_1 distributions for the empty model birdcage coil obtained without the endcap (continuous line) and with the endcap of small diameter ($D_e =$ 76 mm, circles), medium diameter ($D_e =$ 304 mm, triangles), and large diameter ($D_e =$ 375 mm, squares). The endcaps were simulated with d = 12.5 mm. **b**: Experimentally measured *z*-direction RF B_1 distributions for the birdcage coil obtained without the endcap (continuous line) and with the endcap of small diameter ($D_e =$ 152 mm, circles), medium diameter ($D_e =$ 298 mm, triangles), and large diameter ($D_e =$ 375 mm, squares). Data were obtained for the birdcage coil loaded with the oil phantom and the endcap prototype positioned at d = 16 mm.



FIG. 5. **a:** Numerically simulated B_1 gain (%) vs. endcap diameter at several axial positions: z = 0 cm (circles); z = -1 cm (triangles); z = -3 cm (squares); and z = -5 cm (diamonds). Data were simulated for the case of the empty model birdcage coil with the endcap positioned at d = 12.5 mm. **b:** Experimentally measured B_1 gain vs. endcap diameter at several axial positions: z = 0 cm (circles); z = -0.5 cm (triangles); z = -2 cm (squares); and z = -5 cm (diamonds). Data were obtained for the birdcage coil loaded with the oil phantom and with the endcap positioned at d = 16 mm.

of 76–780 mm (with d = 12.5 mm), we found that practically the same optimum ratio $D_{\rm e}/D_{\rm c}$ was also obtained (data not shown). These findings suggest that this optimum ratio of endcap-to-birdcage coil diameter (rather than the expected relationship $D_{\rm e} > D_{\rm c}$) might be explained as a consequence of the "spurious" B_1 field contribution due to the end-ring toward the service end of the birdcage. In fact, previous numerical and analytical calculations (21,22) have shown that the presence of the end-ring does significantly modify the B_1 distribution along the z-direction. We believe that the optimized endcap, positioned at a given distance, produces an end-ring current image that partially counterbalances the B_1 field contribution of the end-ring, thus improving the total B_1 field distribution along the coil axis. However, at present we are unable to predict whether the optimum ratio $D_{\rm e}/D_{\rm c}$ is valid for other field strengths.

In Vivo MRI Results

The sagittal gradient-echo images of Fig. 7 show that in the human head an improved MRI signal amplitude toward the service end was obtained with the slotted endcap



FIG. 6. Numerically simulated B_1 gain (%) vs. endcap diameter at several endcap distances: d = 2.5 mm (circles); d = 12.5 mm (squares); and d = 22.5 mm (diamonds). Data were simulated for the case of the empty model birdcage coil.

present ($D_e = 304 \text{ mm}$, d = 12 mm). MRI signal plots along an axial line of the sagittal image are reported in Fig. 8. Without the endcap the signal amplitude toward the service end was estimated to be 55% of the value at the center of the coil. The optimized slotted endcap gives a significantly improved signal homogeneity along the axial direction, yielding an improvement in signal amplitude to about 85% of maximum toward the service end.

CONCLUSIONS

FDTD analysis and workbench and MRI techniques were used to map the RF B_1 distribution along the coil axis of birdcage coils with and without an endcap. Without an endcap the measured RF B_1 field along the coil axis showed a severe inhomogeneity $(B_1$ at the service end being 45% of maximum). With a detached endcap ($D_e =$ 375 mm) at d = 4 mm from the RF shield, the B_1 amplitude at the service end increases to within 90% of its maximum value. As expected, the B_1 field on the patient side of the coil was practically unaffected by the presence of the endcap. A good agreement was obtained between FDTD and MRI data. Surprisingly, from both the simulation and experimental results we found that there is an optimum ratio, $D_{\rm e}/D_{\rm c} \approx 1.08$, that gives the best B_1 homogeneity along the coil axis (with d = 2.5, 12.5, and 22.5 mm), rather than the anticipated condition, $D_{\rm e} > D_{\rm c}$. From FDTD simulations we also found that the same optimum ratio $D_{\rm e}/D_{\rm c}$ was obtained for the *unshielded* birdcage coil (with d = 12.5 mm). To the best of our knowledge, this optimal endcap design has not been previously reported. In the human head the detached slotted endcap ($D_{e} = 304 \text{ mm}$, d = 12 mm) improves the MRI signal amplitude at the service end from 55% to 85% of maximum. Under these conditions, the small detuning of the birdcage coil due to the presence of the slotted endcap is within the experimentally achievable tuning range.



FIG. 7. 3 T sagittal gradient-echo MRI images (TE = 6 ms, TR = 60 ms, flip angle = 20°, FOV = 35 cm \times 35 cm, slice thickness = 10 mm, 256 \times 256 pixels) of the human head obtained (**a**) without the endcap and (**b**) with the slotted endcap (D_e = 304 mm, d = 12 mm).



FIG. 8. Measured *z*-direction (inferior–superior) MRI signal amplitude in the human head without the endcap (circles) and with the slotted endcap (diamonds) ($D_e = 304$ mm, d = 12 mm).

In conclusion, we have investigated numerically and experimentally the use of optimized detachable endcaps for 3 T birdcage coils. Such designs significantly improve the RF B_1 distribution of birdcage coils, and could prove useful in situations wherein flexibility in access to the RF coil is required.

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