Endourethral MRI

Harald H. Quick,1 Jean-Michel Serfaty,1 Harpreet K. Pannu,1 Rene Genadry,2 Christopher J. Yeung,3 and Ergin Atalar1,3*

Although high-resolution MRI with phased array pelvic, endo-rectal, and endovaginal coils has dramatically enhanced the ability to visualize abnormalities of the female urethra and periurethral tissues, controversy still remains about the anatomy of this region. This study introduces an endourethral approach for ultra-high-resolution MRI of the female urethra and the periurethral tissues. To this end, two different radiofrequency (RF) receiver coil designs for an endourethral insertion have been developed: a single-loop coil and a phased array/quadrature coil. Both designs feature a flexible coil circuit, small loss tuning and matching directly at the coil, active decoupling, and the integration of a λ/4 coaxial choke to decrease unbalanced currents and limit potential RF heating effects. Effective reduction of the mutual inductance between the two coils of the phased array design was achieved by introducing a metallic “paddle” to steer the flux between the coils. The performance of the coils has been evaluated in female human cadaver studies and in an in vivo pig experiment. The novel endourethral approach enabled a dramatic increase of the signal-to-noise ratio (SNR) at the region of interest (ROI). High-resolution MR images of the female urethra have been acquired with a spatial resolution down to 78 × 78 μm. Histologic correlation was achieved for the MR images generated. The achieved high local SNR and resulting high spatial resolution will add valuable information to the discussion of female urethral anatomy. Magn Reson Med 45:138–146, 2001. © 2001 Wiley-Liss, Inc.

Key words: female urethra; endourethral MRI; high-resolution; coaxial choke; decoupling paddle

Detailed anatomic information about the female urethra and pelvic floor is currently unavailable by any diagnostic imaging technique. This information would provide a significant contribution to the understanding of urinary incontinence and other urethral abnormalities, and would be a valuable aid in surgical planning and therapeutic regimens (1). Although normal female pelvic anatomy, as well as pelvic abnormalities, have been reported with the help of body and pelvic phased-array coil MRI (2–5), controversy (2–10) remains about the nature of the urethral layers as demonstrated with MRI. The currently available spatial resolution in MRI of the urethra has not been sufficient to provide an unambiguous correlation of the urethral layers, mucosa, submucosa, smooth muscle, and striated muscle to anatomy (4–6).

The advent of intracavitary receiver coils for high-resolution clinical imaging of the prostate (12,13), and of the uterine cervix (14) has laid the groundwork for endoluminal coil approaches that could image the pelvic floor with increased spatial resolution compared to images acquired with the body coil. Driven by the motivation to further increase the SNR at the ROI, some authors have reported the value of these intracavitary coils in the detailed demonstration of the female pelvic anatomy and abnormalities using both a transrectal imaging approach (1,15) and a vaginal approach to imaging (11,16,17). Although MRI as an imaging modality remains unchallenged in the high soft-tissue contrast needed to study the muscles and ligaments of the pelvic organs, clinical and radiological evaluation of this area remains difficult.

Inspired by the expertise gained in intravascular applications of single-loop receiver coils for high-resolution MRI of vascular walls (18–22), prototype endoluminal MR receiver coils were designed to be inserted into the ROI, the female urethra. Two different coil designs for an endourethral approach are presented: a single-loop coil and a phased array/quadrature coil. A λ/4 coaxial choke was implemented into the design of both coils to reduce potential RF heating effects and to improve coil performance. The implementation of a metallic “paddle” into the phased array/quadrature coil provided effective reduction of the mutual inductance between the two coils of that design. The design process of the coils is described. Heating experiments were performed to ensure the safety of the intraluminal devices. The coil performance has been tested in high-resolution urethral imaging of a formalin-fixed pelvis and of six fresh human female cadavers. The in vivo imaging performance of the coils was evaluated in the urethra of a female pig. An image intensity correction (ICC) algorithm was applied to compensate for the $B_1$ signal variation of the endoluminal coils across the small field-of-views (FOVs) being used. The MR findings were correlated with histology.

MATERIALS AND METHODS

Urethral Coil Design

Based on the constraints and requirements of an endourethral approach for MRI, the outer diameter of the intraluminal coils was limited to 15 F (5 mm), which is comparable to the diameter of a conventional Foley catheter. The coils were of cylindrical geometry and had a length of 50 mm. Thus, the entire length of the urethra, which is (according to values in the literature) in the range of 32–40 mm (11,23), can be covered with the highest sensitivity and homogeneity of the coil. The imaging loop was housed in a biocompatible polymeric tubing with a 5.0 mm outer diameter (OD) and a 4-mm inner diameter (ID). The imaging loop was mounted onto another tubing that added

---

1Department of Radiology, Johns Hopkins University, Baltimore, Maryland.
2Department of Obstetrics and Gynecology, Johns Hopkins University, Baltimore, Maryland.
3Department of Biomedical Engineering, Johns Hopkins University, Baltimore, Maryland.
Harald H. Quick, M.S., is now at the Department of Radiology, University Hospital Essen, Essen, Germany. E-mail: HHHQuick@uni-essen.de
*Correspondence to: Ergin Atalar, Ph.D., Department of Radiology, Johns Hopkins University, Outpatient Center, Room 4241, 601 North Caroline St., Baltimore, MD 21287-0815. E-mail: Eatalar@mri.jhu.edu

Received 20 December 1999; revised 14 August 2000; accepted 18 August 2000.

© 2001 Wiley-Liss, Inc.
suitable to be semi-flexible and to give reproducible results in manufacturing. The conductor separation in its final assembly was determined to be 4 mm. Tuning and matching capacitors were soldered directly onto the surface of the proximal end of the circuit. The coils were tuned to 63.86 MHz and matched to 50 Ohms as they were inserted into a 4.5-l polyethylene container (150 $\times$ 150 $\times$ 200 mm) filled with 0.9% NaCl solution to approximate the loading conditions of a coil inserted into a patient. SNR performance and signal homogeneity tests were performed with the coil inserted in this phantom.

The value of the capacitors was found to be 300 pF for the parallel capacitor, $C_p$, and 95 pF for the series capacitors, $C_s$. Nonmagnetic “case A-size” high-quality ceramic chip capacitors were employed (American Technical Ceramics ATC, Huntington Station, NY). A nonmagnetic small-diameter 50 Ohms coaxial cable with 1.2 mm OD (K 01152-07, Huber & Suhner, Herisau, Switzerland) was used to conduct the signal from the tuning/matching electronics to the surface coil port of the scanner.

To detune the endoluminal receive coil during transmission with the scanner body coil, a conventional active decoupling network was implemented to avoid RF field concentrations at the coil conductors. This was achieved by placing a PIN diode (Type 7204) parallel to the tuning/matching capacitor, $C_p$, between the central conductor and the ground of the coaxial cable. The PIN diode was placed in the coaxial cable at a critical distance ($l = 70$ mm) away from the capacitor, $C_p$. Thus, the inductance, $L_{diss}$, of the coaxial cable and the capacitance, $C_p$, established a resonant circuit with $\omega_0 = 63.87$ MHz when the diode was switched on. During transmission of RF pulses with the body coil, a triggered direct current (DC) voltage at the scanner surface coil port actively switches on the PIN diode. The impedance seen by the receive coil becomes high and detunes this coil during transmission. During receive mode, the diode is opened by an inverse DC voltage provided by the scanner, setting the endoluminal receive coil back to resonance.

A balun circuit was designed by mounting a silver-plated copper braid, 1.59 mm in OD, 19 AWG (Alpha Wire Company, Elizabeth, NJ) with heatshrink tubing onto the coaxial cable. The braiding was connected to the shield of the coaxial cable at the miniature RF plug (MMCX-50-1-2/III; Huber & Suhner, Herisau, Switzerland). The electrical length of the braiding was chosen to form a quarter wavelength, which transforms the short at the RF plug to high impedance at the cable coil conjunction (24). Unbalanced currents are prevented from setting up a resonance on the coil, thus reducing heating effects (25–27). The braiding was constructed to extend over the coil electronics and shield them. In this way the noise performance of the coil was improved, and signal inhomogeneities generated by the entire assembly acting as a rod antenna were reduced. The quarter wavelength of the choke in this configuration was determined to be $l = 910$ mm. The electrical schematic of the entire single-loop coil assembly can be seen in Fig. 2a.

Phased Array/Quadrature Coil

The phased array/quadrature coil assembly is based on the single-loop design described above. Here, two flexible coil circuits were placed orthogonal to each other in the polyamide tubing. Each coil was separately tuned, matched, decoupled, and connected to a separate coaxial line. The values of the capacitors, $C_p$ and $C_s$, as well as the length of the coaxial line between $C_p$ and the PIN diode, were the same as for the single-loop coil.

The two separate coaxial lines were placed inside a common $\lambda/4$ braiding that was held in place around the cables using heatshrink tubing. At the proximal end (i.e., the preamplifier end) of the coil assembly, the braiding was connected to the outer conductors of the coaxial lines at the RF plugs and, therefore, to common ground. The distal end of the braiding extended over the coils’ tuning and matching capacitors. The quarter wavelength of the choke in this configuration was determined to be $l = 930$ mm. The electrical schematic of the entire phased array/quadrature coil assembly can be seen in Fig. 2b.

Minimization of the mutual inductance between the two coils was achieved by inserting a metallic decoupling pad-
dle between the coils from the distal end of the assembly. The use of metal sheets designed to steer the magnetic flux was first described by Bloch, Hansen, and Packard in 1946 (28). The shape of the paddle used in this study was modified and consists of non-magnetic braiding with a rectangular cross-section ($3 \times 1$ mm). The paddle extends over the full length of the coils (50 mm). Twisting the paddle along its longitudinal axis coaxial between the two coils steers the magnetic flux lines and can therefore influence the coupling between the coils in a very efficient manner. The $45^\circ$ position of the paddle relative to both coils was found to result in the most effective isolation of both coils. The mechanical assembly of the paddle relative to the coils is shown in detail in Fig. 3.

Finally, the two coils of the phased array/quadrature design can be combined using a quadrature hybrid coupler for use as a quadrature surface coil; alternatively, the two coils can be used separately with a multicoil system as phased-array coils, without the quadrature hybrid coupler. For our experiments, a dual phased array connection method provided the increase in SNR that theoretically can be achieved with a quadrature configuration. In contrast to a quadrature configuration, however, this approach allowed imaging at any oblique orientation of the coil relative to the static magnetic field $B_0$ without a loss in signal intensity (29).

**Heating Experiments**

Heating experiments were performed with the single-loop coil inserted into an agar gel (DIFCO Laboratories, Franklin Lakes, NJ) phantom doped with 0.9% NaCl to mimic the electrical properties of the body. The conductivity of the gel was determined to be 0.8 S/m at 64 MHz, a median value in the range of human tissues (30). Temperature was monitored with an eight-channel fluoroptic thermometer (UMI 8; FISO Technologies, Inc., Quebec, Canada), by placing five sensors along critical locations of the coil and a sixth sensor at the distal opening of the coaxial choke (Fig. 4). The specific absorption rate (SAR) was calculated by finding the initial slope of the temperature rise ($dT/dt$) and multiplying by the specific heat capacity of the agar gel $C = 4180$ J/kg. Scanning was performed on a 1.5T Signa...
LX (GE Medical Systems, Milwaukee, WI), transmitting with the body coil. The scanning parameters were: fast spoiled gradient (FSPGR) echo sequence (FSPGR: axial sections, flip angle 122°, TR/TE 5.8/1.9 msec, FOV 480 × 3480 mm, matrix 256 × 128, 113 NEX, BW 31.25 kHz, 4 slices each 10 mm, 10 mm spacing, imaging time 9:56 min). Setting the weight to 100 kg, the scanning parameters resulted in a peak SAR of 3.96 W/kg and a mean SAR of 1.98 W/kg, as calculated by the scanner software. The phantom and cables were kept parallel to the main axis of the scanner bore. The cables were elevated to the isocenter of the magnet and moved laterally until they were as close to the bore wall as possible (8 cm), as constrained by the size of the phantom. This position gave near maximal heating (25). For each tested position the coil was additionally disconnected to simulate the failure mode with disabled detuning of the endoluminal coil.

In Vitro Experiments

Imaging was performed using the body coil of the scanner as an RF transmitter and the endourethral coils as receive-only probes. For assessing the signal characteristics of the antennas, proton density (PD)-weighted axial and coronal images were acquired with a fast spin-echo (FSE) sequence (PD-FSE: 8 echoes, TR/TE = 2000/14.5 msec, BW = 15.6 kHz, 6 NEX). Axial 3-mm sections with a 30 × 30 mm FOV and a 256 × 128 matrix (Fig. 5), and coronal 2-mm sections with an 80 × 80 mm FOV and a 256 × 128 matrix (data not shown) were collected in 3:24 min.

The imaging performance of the single-loop coil was evaluated in one formalin-fixed cadaver as well as in six freshly harvested female human cadaver pelvices. The coil was introduced into the urethra with the tip entering the bladder approximately 5 mm. This position ensured that the entire urethra and the internal sphincter were covered by the highest signal intensity and homogeneity of the coil. For localization purposes, a 5-inch × 9-inch rectangular-shaped surface coil was employed from outside the body to give gross orientation of the small pelvis anatomy. This coil was disabled for subsequent high-resolution imaging with the endourethral coils. Following double oblique sagittal and coronal FSPGR localizer imaging with large FOVs, high-resolution PD and $T_2$-weighted imaging of the entire urethra was acquired in a plane axial oblique to the urethra coil. The spatial in-plane resolution was progressively increased by consecutive reduction of the FOV from 40 × 40 mm to 30 × 30 mm and finally to 20 × 20 mm. All images were zero interpolated (ZIP) in the in-plane direction to a 512 × 512 matrix to improve the display quality. Specific imaging parameters are given in the legends to Figs. 6–8.

In Vivo Experiments

The in vivo imaging performance of the single-loop coil was evaluated in the urethra of a healthy, fully anesthetized female pig weighing 55 kg. The experiment was conducted in accordance with all regulations set forth by institutional and governmental agencies. The imaging coil was inserted into the urethra with the pig lying in the supine position. Following double oblique sagittal and coronal FSPGR localizer imaging with large FOVs, high-resolution PD and $T_2$-weighted imaging of the entire urethra was acquired in a plane axial oblique to the urethra coil. Specific imaging parameters are given in the legend to Fig. 9.

Image Intensity Correction

The signal intensity of images acquired with small endoluminal imaging probes is highly non-uniform due to the $B_1$ inhomogeneity of the pick-up device (31). High sensitivity of the receiver coil in its immediate vicinity and especially close to the coil conductors leads to a signal saturation in these areas, whereas the signal drops off with roughly $1/r^2$ with increasing radius, $r$, away from the coil. It is difficult, if not impossible, to see all the parts of the images at any one time by adjusting the contrast and brightness of the images. A solution to this problem is to calculate a uniform signal intensity image by dividing the image intensity by the corresponding sensitivity map of the coil. In this case,
noise on the images is not uniform and will increase toward the periphery of the image, but a single brightness and contrast level can be used for a convenient display of the central parts of the image (20). The applied display method reads an image, derives the main magnetic field orientation with respect to the imaging plane, and assumes that the coil is perpendicular to that plane. Coil positions and angles can be entered interactively.

Histology
Following high-resolution MRI, the endourethral imaging coil was removed and a Foley catheter 15 F (5 mm) in diameter, and marked with an mm-scale, was inserted into the urethra to facilitate the matching process between MR images and histologic specimens. The balloon of the catheter was blocked inside the bladder with 10 ml of water. The cadaver was subsequently imaged with the rectangular surface coil to document the position of the Foley catheter running through the urethra. Histologic correlation of the MR images was achieved by removing the urethra en bloc from the unfixed cadavers. The specimens were fixed by suspension in 10% buffered formalin. During the fixation process, the catheter preserved the lumen of the urethra. The mm-scale on the catheter helped to correlate the specimens with the position of the acquired MR images. The specimens were embedded in paraffin and

---

**FIG. 6.** Localizer images, (a) axial and (b) sagittal, of a female human cadaver pelvis acquired with the single-loop endourethral receive coil inserted into the urethra (200 × 200 mm image portions are shown). The images were acquired with the body coil of the scanner. No increased signal around the urethra coil could be observed, indicating good performance of the coil’s decoupling circuit. Sequence parameters were: fast spoiled gradient echo sequence (FSPGR: flip angle 60°, TR/TE 150/5.4 msec, FOV 360 × 360 mm, matrix 256 × 160, 1 NEX, BW 31.25 kHz, slice 5 mm). Letters mark the pubic bone (PB), rectum (R), and vagina (V). The distal tip of the coil is inserted into the bladder approximately 5 mm (arrow).

**FIG. 7.** High-resolution images acquired in an axial plane with the single-loop coil inside the urethra of a human cadaver (see also Fig. 6). Sequence parameters were: T2-weighted fast spin-echo sequence: (FSE: 8 echoes, TR/TE 2000/78 eff msec, FOV 30 × 30 mm, matrix 320 × 192, 6 NEX, BW 15.6 kHz, slice 1.5 mm, spacing 1.0 mm, imaging time 4:48 min for 13 slices). The upper row shows images acquired at the level of the distal end of the urethra (a), the mid-urethra (b and c), and the bladder neck (d). The bottom row shows the corresponding images after image intensity correction. Letters mark the pubic bone (PB, no signal) and the vagina (V, bright band) (f). Three layers of the mid-urethra can be depicted (g): the hypointense submucosa (SM), the striated muscle (STM) appearing bright, and the hypointense smooth muscle (SMM). Fluid accumulations (FL) in the urethra (f and g) and in the bladder neck (h) appear with bright signal. Note the increased contrast following image intensity correction.
5-μm-thick axial sections were cut and stained with Masson trichrome staining on glass slides.

RESULTS

Urethral Coil Design

The quality factor, $Q$, of the coils loaded in NaCl solution was determined to be 67 for the single-loop, and 74 and 80 for the phased array/quadrature coils, respectively. The impedance of the quarter wavelength coaxial chokes was determined to be 730 Ohms for the single-loop coil design and 750 Ohms for the quadrature/phased array coil design. The isolation paddle enabled the minimization of the mutual inductance between the two coils in a very effective manner. Twisting the paddle close to a 45° position was found to result in the highest isolation of both coils. An isolation of greater than 50 dB was obtained with negligible loss of performance. The decoupling was not significantly affected when the loading of the coil was changed. Thus, no readjustment of the position of the paddle was necessary.

For each position of the coil/phantom in the scanner, the measured SAR increases from sensor 1 to sensor 5 (from the inside of the phantom to the edge of the phantom) (Table 1). The SAR increases when the coil is disconnected from the scanner surface coil port (when the decoupling of the receive coil is disabled (“failure mode”)) (Table 1, position b). The SAR increases when the phantom/coil is moved toward the scanner bore (closer to the RF transmitting body coil) (Table 1, position c). Positioning the phantom/coil close to the wall of the scanner bore and disconnecting the coil from the surface coil port (Table 1, position d) gives maximal SAR (5.32 W/kg) at the proximal edge (sensor 5) of the imaging coil (Fig. 4).

Introduced into saline solution, the single-loop coil showed good axial signal homogeneity over the full circumference of the coil (Fig. 5a). The near-field of the coil reveals two high signal spots directly at the coil conductors. The phased array/quadrature receiver coil initially showed coupling between the two coils, resulting in decreased signal penetration depth and homogeneity (Fig. 5b). Inserting the decoupling paddle results in dramati-
focus of this study was on the generated T2 signal. On proton density (PD) and full length of the urethra with high and homogeneous coverage of the longitudinal axis of the coil ensured coverage of the urethra in its full length. The signal characteristics along the coil/phantom in the scanner were: a) middle of bore, coil connected; b) middle of bore, coil disconnected, i.e., decoupling disabled; c) close to scanner bore, coil connected; and d) close to scanner bore, decoupling disabled. Positions of the sensors relative to the coil are shown in Fig. 4.

**Cadaver Experiments**

The acquisition of axial and sagittal localizer images with the body coil while the single-loop urethral coil was inserted into the urethra revealed no increased signal adjacent to the endourethral coil. This indicates that the decoupling circuit reliably detuned the receiver coil during coupling, with the tip of the coil being inserted into the bladder and represent most likely submucosal vascular plexus and glands. Image contrast increased by the removal of the signal’s B1 signal profile, resulting in better visualization of the different urethral wall layers (Fig. 7g). The fluid-filled urethral lumen had a bright signal on T2-weighted images (Fig. 7g). The mucosa of the urethra was also hyperintense on T2-weighted images and indistinguishable from fluid. Surrounding the lumen, a circumferential hypointense layer was identified that corresponded to the submucosa (SM). A hyperintense thicker layer, corresponding to the smooth muscle (SMM), encircled the urethra. The hypointense layer, surrounding the smooth muscle layer, represented the striated muscle (STM).

Further decrease of the FOV to 20 mm, while increasing the spatial resolution to 78 × 78 μm, enabled the resolution even of single folds of the urethral submucosa (SM) (Fig. 8b). The image intensity correction was again found to be helpful in enhancing image contrast and in resolving structures adjacent to the coil.

**In Vivo Experiments**

Heavily T2-weighted axial images of the midurethra, acquired with the single-loop coil in the urethra of a pig, revealed improved tissue contrast when compared to the images of the cadaver experiments (Fig. 9). Despite high in-plane spatial resolution of 78 × 78 μm and image acquisition times of several minutes, the images acquired in vivo were of high quality and free of blurring and motion artifacts.

**DISCUSSION**

Evaluation of the female urethra presents challenging clinical and radiological problems. Opinions about the nature of the layers of the urethra as demonstrated with current MRI techniques (including the transrectal and the endovaginal approach for coil placement) differ in the literature (11). The ability to image the urethra at higher resolution than is currently available is mandatory for the investigation of incontinence and other urethral abnormalities (6). In this regard, the development of MR receive coils for the novel endourethral approach introduced in this study enabled the acquisition of ultra-high-resolution MR images of the female urethra with a spatial in-plane resolution down to 78 × 78 μm, acquired within 8:32 min.

Requirements for the design of an endourethral RF receive coil include: 1) a device diameter small enough to fit the lumen of the urethra, 2) longitudinal signal coverage of the length of the urethra, 3) minimized radial sensitivity.

---

**Table 1**

<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>a)</td>
<td>0.63</td>
<td>0.67</td>
<td>1.05</td>
<td>1.26</td>
<td>1.68</td>
<td>1.54</td>
</tr>
<tr>
<td>b)</td>
<td>1.12</td>
<td>1.12</td>
<td>1.75</td>
<td>1.68</td>
<td>1.82</td>
<td>1.70</td>
</tr>
<tr>
<td>c)</td>
<td>1.68</td>
<td>1.4</td>
<td>1.82</td>
<td>2.31</td>
<td>2.45</td>
<td>2.07</td>
</tr>
<tr>
<td>d)</td>
<td>3.78</td>
<td>3.78</td>
<td>4.34</td>
<td>5.04</td>
<td>5.32</td>
<td>4.32</td>
</tr>
</tbody>
</table>

The imaging region of the coil as well as the mouth of the coaxial choke were inserted into an agar-gel approximately 75 mm. Positions of the coil/phantom in the scanner were: a) middle of bore, coil connected; b) middle of bore, coil disconnected, i.e., decoupling disabled; c) close to scanner bore, coil connected; and d) close to scanner bore, decoupling disabled. Positions of the sensors relative to the coil are shown in Fig. 4.
fall-off to the coil to improve the penetration depth, 4) a homogeneous response to radially equidistant objects, and 5) a homogeneous response to axially equidistant objects. In addition, the safety of the device for humans must be considered. An important safety concern associated with the incorporation of an RF antenna into the human body is the possibility of localized RF heating caused by the coil or the coaxial cable connecting the coil to the scanner (25,27). While the small-area loop coils in this investigation are not considered a high risk in terms of RF heating, the overall design with a coaxial cable connecting the coil to the scanner could potentially pick up RF energy, leading to tissue heating at the coil. The implementation of coaxial chokes into the coil design has been shown to be an effective means of reducing RF heating in intravascular MR applications (26,27). None of the tested cases resulted in excessive RF energy deposition beyond the local SAR limits (Table 1) based on the current International Electrochemical Commission (IEC) 60601-2-33 standard: 8 W/kg in any gram of tissue in the head or torso for 5 min, or 12 W/kg in any gram of tissue in the extremities for 5 min (32). The prototype coils investigated in this study were therefore considered to be safe by a university’s (Johns Hopkins University) institutional review board for potential use in humans.

In addition to the reduction of heating effects, the implementation of a coaxial choke into the overall design resulted in an improvement of coil homogeneity and thus image quality. Two effects were accountable. First, the braiding of the choke was extended over the coil tuning and matching capacitors, shielding the resonant components from exterior conducting structures and thus potentially reducing dielectric losses. Second, the choke limitted unbalanced currents on the outer shield of the coaxial cable, which potentially can interact with the $B_1$ field of the coil, diminishing its homogeneity.

The use of a metallic “paddle” to steer the magnetic flux of a coil as a means to isolate two adjacent coils from each other was first published in the literature in 1946 (28) and was then reconsidered by others (33,34). In these studies, however, a rather small paddle was used to decouple large volume coils from one another in an attempt to minimize distortions of the $B_1$ field homogeneity to the inside of the coil. The application of such a paddle to an “inside out” design of endoluminal coils, as described in this work, enabled the paddle to be inserted into the most sensitive region between the coils. Furthermore, the paddle could be designed larger relative to the size of the coil, providing a very effective means of coil isolation, virtually without affecting the $B_1$ field toward the outside of the coil (Fig. 5c). No additional electronics were required to achieve an isolation of about 50 dB. Image quality was markedly improved by minimizing the mutual inductance between the coils (Fig. 5b and c). Without the paddle inserted, the signal performance of the phased array coil was inferior compared to that of the single-loop coil (Fig. 5a and b) due to the interaction between the two coils. Changing the loading of the coil after adjustment of the paddle did not significantly affect the decoupling, so paddle readjustment was not required.

Reflecting the more complex mechanical design with two independently tuned and matched flexible coil circuits (and therefore its higher susceptibility to mechanical stress), the phased array coil configuration failed during the cadaver experiments and was not investigated further in this study. The better performance of this design compared to the single-loop coil with regard to axial signal homogeneity, SNR, and signal penetration depth into the tissue (Fig. 5), however, justifies further study of this concept for potential clinical applications.

The small diameter and the semi-flexible, cylindrical design of the coils enabled easy insertion into the cadaveric female urethra. In addition, the design allowed stable positioning of the coil within the urethra relative to the periurethral tissues. The length of the coils was sufficient to cover the full length of the urethra with high and homogeneous signal. The conductor separation of 4 mm provided sufficient signal penetration depth into the tissue to image the full axial extent of the urethra which is, according to values in the literature, in the range of 10–15 mm (11,23), with high spatial resolution. Moreover, signal beyond the urethra enabled the examination of periurethral tissues. Although the coils were comparable in size and shape to a conventional Foley catheter, and tolerance by the patient is thus anticipated, the introduction of a semi-flexible device with an OD of 15 F into the urethra limits, if not precludes, the performance of dynamic studies (35,36). Further improvement of the coil design with regard to flexibility will, it is hoped, enable the visualization of urethral displacement during straining, which would add valuable dynamic information to the achieved static high-resolution imaging of this region.

The proposed image intensity correction algorithm applies a theoretical model of the magnetic field distribution and requires interactive positioning of the coil conductors from the user, thus potentially making the resulting corrected images subject to systematic or user-dependent errors. The application of the IIC, however, reliably removed signal inhomogeneities and hot spots due to the coil conductors, thus facilitating the visualization of structures adjacent to the coil (Figs. 7e-h, 8b, and 9b). Tissue contrast was successfully regained after image correction. Interactive displacement of the coil conductors did not significantly affect the outcome. The IIC is therefore considered to be robust for the endourethral application. Rather than applying a theoretical algorithm to a “real” in vivo situation, the algorithm proposed by Nanz et al. (31) employs two sets of images acquired in vivo to extract the “real-life” $B_1$ signal distribution, thus correcting for potential systematic errors that could cause deviations of the theoretically assumed $B_1$ signal distribution, i.e., coupling artifacts and angulation of the coil in the main magnetic field $B_0$.

The acquired $T_2$-weighted axial images enabled the visualization of the different layers of the urethra at high resolution (Figs. 7 and 8). Our findings are in accordance with the results of a previous in vitro investigation (6). Although the cited study focused on high-resolution MRI of female urethral specimens and achieved excellent histologic correlation, the applied volume wrist coil is not applicable to in vivo image acquisition.

As demonstrated in the pig experiment, in vivo imaging with the endourethral approach provided high-quality images free of blurring and motion artifacts despite high in-plane resolution and image acquisition times of several minutes.
In conclusion, the experimental data of this study have proven that intrarethral MRI can provide very high-resolution images of the female urethra and surrounding tissues. It may therefore become an important adjunct to urography and functional methods for the diagnostic evaluation of incontinence, thus contributing to surgical planning and facilitation of surgical correction.

ACKNOWLEDGMENTS

The authors thank Mary McAllister, Tammy L. Oreskovic, M.S., and Ronald Ouwerkerk, Ph.D., for their help in manuscript preparation. The authors are grateful to Perry Karmarkar, M.S., of Surgi-Vision, Inc., for providing the flexible coil circuits and fabricating the prototype coil used in the in vivo pig experiments. Ergin Atalar is a co-founder and consultant of Surgi-Vision, Inc.

REFERENCES

26. Ladd ME, Quick HH. A 0.7 mm triaxial cable for significantly reducing RF heating in interventional MR. In: Proceedings of the 7th Annual Meeting of ISMRM, Philadelphia, 1999. p 104.