Novel transmitter designs for magnetic resonance imaging

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6

General Discussion and Future Developments

This is a concluding chapter wherein the scientific and technical implications for society of the research findings of this thesis are discussed, possible future developments are also identified.

6.1. GENERAL DISCUSSION

6.1.1. DISCUSSION ON DIELECTRIC RESONATORS

Dielectric resonators are promising new RF coils for human magnetic resonance imaging.Whilst they have been previously described in the literature for human imaging by Wen [37], through the presently described work we have systematically developed dielectric resonators to a stage that they are now usable by other researchers and RF engineers in the field of human ultra-high field magnetic resonance imaging.

Through the research described in chapter 2, it was for the first time demonstrated that dielectric resonators can be used as effective transmit and receive coils for high field magnetic resonance imaging. The use of the HEM mode and thus the possibility to use quadrature feeding of the coil offers an increase in efficiency of $\sqrt{2}$ over the linear drive mode. Also, the coaxial orientation of the HEM resonator with respect to B_0 is advantageous when building dielectric resonators conformal to the human body.

It was shown that with a very simple and inexpensive design, RF coils that use water as a dielectric can be designed and built. These dielectric coils can be used for human magnetic resonance imaging with a performance very similar to the established birdcage resonator design. This is not surprising if one considers how a birdcage resonator is constructed. It also allows a stable, time-invariant electromagnetic wave to exist in its lumped element network. The higher order modes of the birdcage resonator can be viewed as a small sample of possible higher order modes of a similar sized dielectric resonator. This analogy becomes even more clear when the number of rungs of the birdcage resonator is increased; as the number of rungs approaches infinity, the structure of the bird cage resonator approaches that of a dielectric resonator coil.

One major aspect of RF coil design, besides improving efficiency, is that the specific absorption rate of the human body must be taken into consideration as it imposes conditions for safe operation of the RF coil. In a birdcage resonator, the electric fields which cause tissue heating emanate from the capacitors thereof. The electric field strength is highest around these components. The dielectric resonator has an advantage in that there are no lumped elements and thus the electric field that interacts with the human body is distributed out over a much larger surface area thus potentially lowering the SAR value.

A further advantage of dielectric resonators over birdcage coils is that much smaller length to diameter ratios can be used. This is useful to the coil builder; or access to extremities, shorter coils are desirable as they restrict the motion less and bring more patient comfort.

It has been shown that the earlier published empirical formulas, although derived from dielectrics with a lower permittivity and at an operating frequency in the GHz range, are still quite accurate for water as a dielectric and in the MHz range.

While our first resonator used water as a dielectric, it was desired to use a more common dielectric material (like barium titanate for example): water has a relatively high loss tangent when compared to barium titanate or perovskites in general. However, it has to be born in mind that the Q-value of a dielectric resonator is only very high in the unloaded case ($Q_{unloaded} = \frac{1}{tan\delta}$). A very high Q would be favorable in a situation where the dielectric resonator is used as a radiative element, such as as described in chapter 4 for the 7 Tesla cardiac array, in order to radiate power away from the element into the body. If the DRA is to be used similarly to a conventional birdcage or loop gap resonator, as described in chapters 2 and 3, one might choose a lower Q in order to maximize the field within the resonator instead of radiating it away from the coil.

Effective detuning of an RF coil is needed if the coil is to be run in a transmit or receive only mode as is common nowadays in commercial magnetic resonance imaging scanners for humans. Spin excitation is usually achieved using the large body coil during the transmit phase. The body coil is then detuned and a local receive coil is switched from a detuned to a tuned state. It is thus very important for any new coil design to also have this capability. We have shown in chapter one that it is possible to effectively detune a dielectric resonator with a simple PIN-diode circuit that is very common in RF coil design. The detune capability was a major step forward into using the dielectric resonator technology for human magnetic resonance imaging.

It is also important to be able to fine tune the resonance frequency of a coil. This can be needed in daily practice in order to compensate for different loads but even before that, it is important to have some mechanism to fine tune the resonance frequency of a RF coil on the workbench. This problem becomes even more pertinent if sintered high Q ceramics are used for dielectric resonators. Despite accurate estimation of the mode and thus the needed dielectric constant using electromagnetic simulations, small deviations from the specified dielectric constant and dimensions can sometimes not be avoided due to imperfections in the production process. As the production of high Q ceramics is a lengthy and costly effort, the facility to be able to fine tune a resonator that is "a couple of MHz off" becomes even more important. As we have shown in chapter two, this is possible by simply changing the boundary condition of the dielectric resonator with an inexpensive copper foil. We have shown that by changing the boundary condition, it is even possible to double, and triple tune a cylindrical DRA. This feature could be for example exploited for double tuned water / fluorine imaging.

To be able to design dielectric resonators as effective RF coils, certain formulas are needed that help the design process. These formulas can be obtained from the literature [13, 19, 28, 32, 35] and work well in the frequency regime for current human MRI. Thus, lengthy trial-and-error builds of coils can be avoided. We were able to show that the empirically derived formulas from dielectric resonators in the GHz range and with much lower permittivity are still valid for much larger dielectric resonators that work in the MHz range. We have also shown in chapters 1, 2 and 3 that these dielectric resonators can be designed using numerical simulations. Especially eigenmode analysis proved to be a quick and reliable tool for the design and field pattern prediction of cylindrical DRAs. While similar results can also be obtained with a time-domain simulation approach using the FDTD [31, 36] method, the eigenmode analysis proved to be much faster producing results in minutes as compared to hours with the FDTD method. Our results should encourage other research groups also to carry out investigation into dielectric resonators as electromagnetic simulation software is nowadays widely available among the ultra-high field magnetic resonance imaging laboratories of the world.

6.1.2. Discussion on Plasma based RF Coils

In chapter 4 we have described for the first time a novel MRI coil that uses plasma as a conductor. We were able to show that a plasma based RF coil cannot only be used to transmit and excite the spin system but also that it can be used to receive the magnetic resonance signal.

This truly novel concept of an RF coil for magnetic resonance imaging was realized using inexpensive fluorescent lamps. While those have a too high internal gas pressure to be really power efficient at 300 MHz, they are largely abundant in hardware stores all around the world. Other research groups can easily rebuild our designs to start research on this new resonator technology.

Plasma technology offers unique and attractive features that cannot be achieved with conventional lumped element RF coils or dielectric resonator RF coils. The conductor in a plasma coil is the plasma itself. Plasma can be switched on and off as wished for example for the purposes of detuning for reasons mentioned in the last section. The difference here is that while a lumped element coil such as a birdcage resonator or a dielectric resonator can also be detuned, there will always be a physical conductor left. This is unavoidable and undesirable as it alters the receive field and attenuates the gamma rays in case of multi-modal imaging like PET-MRI. A plasma in contrast, if switched off, is physically no longer present. Only the container of the plasma and the gas it contains is present. This will allow for very compact designs of transmit and receive RF coils that are very close to each other.

We were also able to show that the length of which the plasma ignites inside its tube can be adjusted by the input RF power that is also used for ignition of the plasma. This gives an additional degree of freedom to control the extent of the B_1 field. We showed that a simple surfatron can be built that acts simultaneously as a plasma igniter and at the same time as a coupler for the RF into the plasma antenna.

Our experiments show that the fear of increased noise from the plasma leading to no useful MRI at all is unsubstantiated. In fact in theory even lower noise floors than those of metal antennas should be possible if the right plasma configurations are chosen [2].

The efficiency of all plasma experiments conducted so far was lower than when using a conventional RF coil. One should however not forget that the current state of

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conventional RF coil design is based on research collated over more than 40 years from all around the world. Also, so far, no efficient coupling to the plasma column has been realized in MRI. We thus can hope for increased efficiency from this new promising coil design.

One first step to increase overall efficiency and usability of plasma MRI coils would be to produce custom plasma columns. These could be made from glass but also certain foams like SynFoam [2] can be used to contain a plasma. These new columns or housings can also have an arbitrary shape. Once manufactured they need to be filled with a suitable gas. One option would be to use atmospheric air and then lower the pressure to allow forming of plasma. The density of the plasma needs to be lower than in current fluorescent light tubes. This would allow the plasma frequency to be steered (also known as Langmuir frequency) into a regime interesting for MRI.

The second step to improve overall efficiency and usability of plasma based MRI coils would be to design, built and test improved coupling schemes to the plasma. Plasma physics is not an entirely new field. In fact, plasma physics research in the twentieth century has produced numerous solutions for this problem [4, 6, 26, 27, 30]. These solutions need to be revised and tested to determine whether they are suitable for MRI.

The third step to improve overall efficiency and usability of plasma coils in MRI would be to look into alternative plasma ignition and mean of sustaining the plasma. We so far used only the transmit pulse as an ignition source. This has so far proven to be not very effective. Perhaps much better results could be achieved if the plasma were to be ignited by another mechanism. Also, here the plasma research of the last century is a vast source for inspiration. One example that was recently published is that DC biased plasma columns can be used as antennas [5]. Also, classical literature can give more hints about how plasma behaves in a strong magnetic field. These results have been published [12] and should be studied to gain insights on how to improve plasma based antenna technology in general and plasma base MRI coils in particular.

Overall, we can conclude that plasma based applications in MRI are in their infancy. The field is now open for a lot of new ideas and solutions to be found. These efforts will hopefully help one day to make MRI and human imaging diagnostics of the future better for all mankind.

6.2. FUTURE DEVELOPMENTS

6.2.1. IMPROVED COUPLING SCHEMES FOR DRAS

While we used an inductive coupling scheme for all our dielectric resonators, numerous other coupling schemes are conceivable and have been described in the literature [11, 13, 19, 32]. The direct coaxial feed is a very interesting application and candidate for an improved coupling scheme. With such a feed, the resonator could be made even more compact, as space for critically coupled loops would be no longer needed. Especially if one considers dense radiative arrays that are made up of DRAs similar to the array of chapter 3, feeding of individual elements via a printed circuit board and micro-striplines might be considered. This would permit thinner arrays to be built.

While in chapters 2 and 3 the dielectric resonator was used to surround the object to be imaged, in chapter 4 an array that works based on TE mode resonators was described. In such an array it is desirable to radiate as efficiently as possible the applied power into the human body in order to increase SNR. The efficiency of such an array could be further optimized by choosing a better suited dielectric permittivity for the resonators. It is conceivable to shrink the resonators to such an extent that one single DRA would have dimensions in the order of a few centimeters. Such resonators could then be placed on a pcb board that is equipped with micro-striplines to couple energy into the DRAs. In this way, a dense high channel count array could be produced similar to that described in [19]. With phase steering using PIN-diode based phase shifters one could imagine that the image quality for cardiac imaging at ultra-high fields could be further improved in the future.

In our research we used exclusively cylindrical DRAs. However, one can essentially use any shape of DRA provided that the mode is suitable for the experiment to be carried out. It is thus conceivable to use very high permittivity (>1000) materials [38] that could be manufactured into thin slabs. This way an even thinner array could be realized having a thickness that approaches the thickness of lumped element circuits. Such a thin DRA could then be used as a substrate for other lumped elements coils, for example a receive array. One could use the DRAs for the high power transmit and the lumped element coil array to receive the MR signal. This way ultra-compact transmit/receive arrays could be realized.

6.2.2. ODD RATIO DRAS

Dielectric resonator can be built with odd ratios of height to length when compared to classical MRI coils. For example, in the birdcage resonator, the ratio between length and diameter should be around $\sqrt{2}$. We were able to build such a coil and early results for a simple prototype are given in figure 6.1.

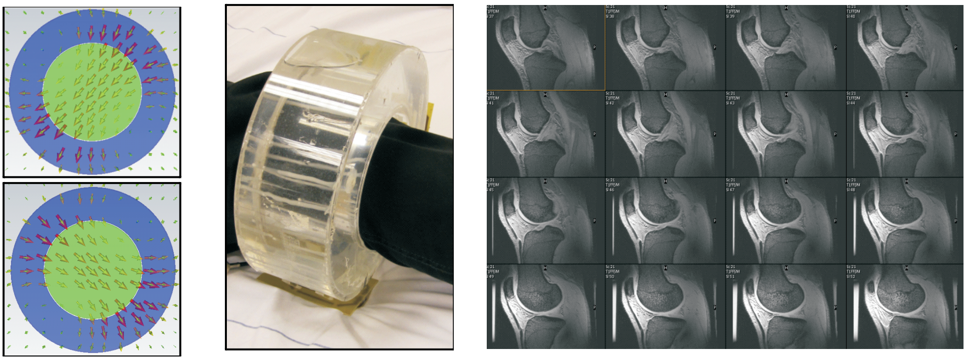

Figure 6.1: Principle of a odd length to diameter ratio dielectric resonator MRI coil: (left)Simulated magnetic field components corresponding to the two frequency degenerate HEM11δ modes resonant at 298.1 MHz. Blue represents the outer annulus filled with water and the green is a homogeneous tissue phantom assigned dielectric properties of tissue. (center) Photograph of the assembled dielectric resonator placed around a volunteer's knee. The two coupling networks can be seen underneath the resonator. (right) A sub-series of adjacent slices from a three-dimensional T1-weighted imaging sequence.

The benefit of such designs could be increased patient comfort while using less space within the MRI bore. Clearly this design needs to be further improved. For example, should the ring be split-able into two half circles, the bottom half could be then put on the patient bed of the MRI system, the knee of a patient would be then placed in the center of this half and thus when the patient would be rested, the top half would be attached. This work shows that odd ratio diameter DRAs are feasible and can be used for MRI as theorized in the discussion of chapter two [3].

6.2.3. HYBRID DRA / PLASMA RF COILS

It is also conceivable to combine dielectric resonators with plasma RF coil technology. One can for example think about nesting plasma columns inside a water based DRA. This would then allow double tuning while perfectly decoupling the two coils from each other. The need for switching circuits will be minimal like shown in chapter 2.

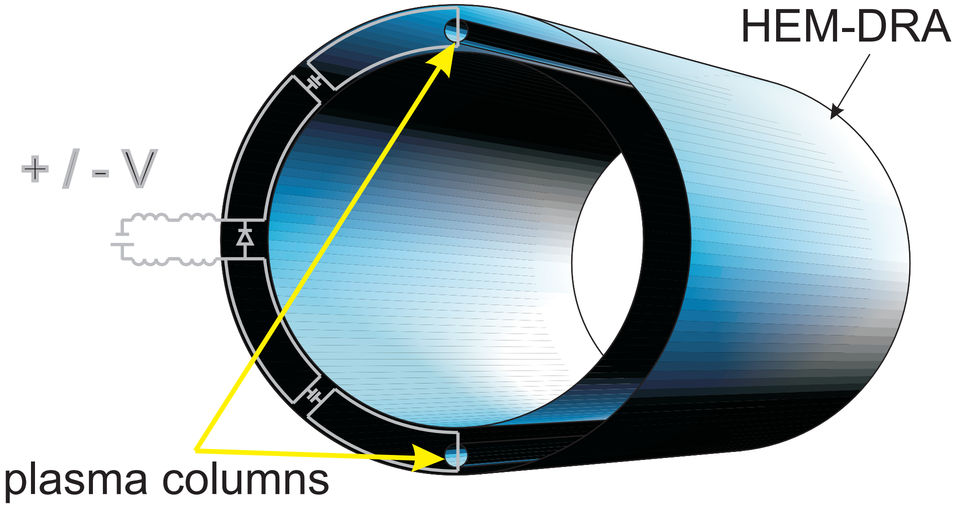

Figure 6.2: Principle of a hybrid plasma / dielectric resonator MRI coil: a DRA using the HEMδ modes is constructed. Within the dielectric resonator plasma columns are nested. In the operation of the DRA for example at proton frequency the plasma is switched off. In order to detune the DRA the plasms is switched to the on stage. Thus, it is also possible to use the plasma columns for excitation of a second frequency, for example phosphor.

We have shown that detuning of a DRA works with the circuit as described in chapter 2. Detuning of the circuit works because the rods shorten the electric field in the HEM resonator. This would also happen if plasma were used as a conductor instead of the copper we used in the experiments of chapter 2. One might speculate that even this simple switching circuit may no longer be needed as the off state of the plasma basically is essentially an open circuit, i.e functioning like the diode in the circuit in figure 6.2.

6.2.4. SILENT PLASMA GRADIENT COILS

We have shown in our work in chapter four that plasma can be used as a conductor in MRI to transmit and receive an MR signal. Another possible application for plasma in MRI would be to use it as a conductor in a gradient coil instead of a copper conductor. Gradient coils have basically been constructed in the same way for the last 40 years: a copper conductor is wound around a stiff glass-fiber carrier tube, water cooling pipes are added and the whole package is then immersed in resin and cured. This is needed to provide the necessary stiffness to counteract the massive Lorenz forces that are present during operation of a gradient coil. These forces are also responsible for the characteristic loud noise and banging as well as vibrations inside MRI system [20]. The noise and banging are one major issue with regards to patient comfort and hearing safety for MRI in its current state [7]. Much effort has been put into reducing these effects including putting the gradient coil in a vacuum vessel [14], active noise control [21, 22], using specialized pulse sequences [9] as well as using swept radio frequency [10]. However, all those efforts do not remove the problem but cure some of the symptoms - the vibrations of the gradient coil are still present. Over time those vibrations can cause massive material stress leading to material failure, in best case only the clamps that hold the gradient c

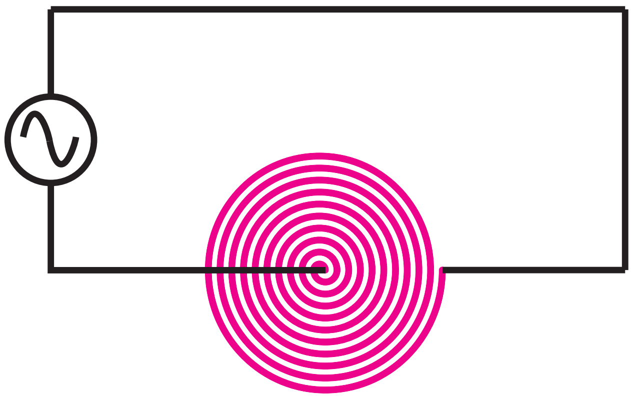

Figure 6.3: Principle of a plasma conductor based gradient coil. The copper conductor is removed and replaced by a plasma conductor.

It might be possible to overcome these unwanted properties of gradient coils if the copper conductor were to be removed and replaced with a plasma guide structure as illustrated in 6.3. Plasma can handle very high powers (for example millions of amperes are conducted to earth in lightning bolts); theseveral hundred amperes needed for a gradient coil nowadays should not be overly problematic. The Lorenz force will still work upon the plasma; however, one can speculate that much less mass will be moved as the density of the plasma is of course much lower than that of copper metal. If the Lorenz force is significant enough to allow for noise and vibration generation, one could use a trick that would not be possible with a metal conductor to still get a truly silent gradient coil: the Lorenz force will produce a pressure inside the containment filament. This overpressure might be conducted out of the plasma coil into another expansion chamber wherein the pressure could be absorbed by a dampening system. One could consider either a form of silencer or even just an open box with a valve. Such a system would have the added advantage that the plasma ceases to conduct if expanded. This would help to ensure that power is only conducted inside of the plasma gradient coil and not in the expansion /silencer box.

In general, two different designs are conceivable:

1. Ignition on demand with coupled power (gradient wave form) feed over a single surfatron
2. Continuous ignition of the plasma and separate feed of power (gradient wave form)

For the first design a high power surfatron would need to be build. One question that comes to mind for configuration is whether plasma over the whole length of the coil could be switched on and off fast enough. If this should prove not to be possible, design number two provides an alternative. Here the plasma could be pumped and thus sustained with a different frequency than the gradient pulse frequency. This has been previously shown to work for plasma antennas in the MHz regime [34, 39].

6.2.5. Nonlinear Plasma Coils for multi-band MRI

In multi-band MRI the time required for acquisition of the MR image is reduced by exciting not only one slice but at least two slices simultaneously [25]. In order to decode the receive signal it is desirable to have clearly identifiable coil sensitivity profiles. Mostcoils nowadays are built from equal coil elements to from an array. If one would use plasma coils as elements, it would be possible to dynamically alter the coil sensitivity profile for each element by adjusting the amount of ionization of the plasma, for example by driving a single element with more or less power. If the plasma coil were to be used also as a transmit coil, one could dynamically alter the B_1^+ field and thus add a degree of freedom. This behavior was tested experimentally;the results are shown in 6.4.

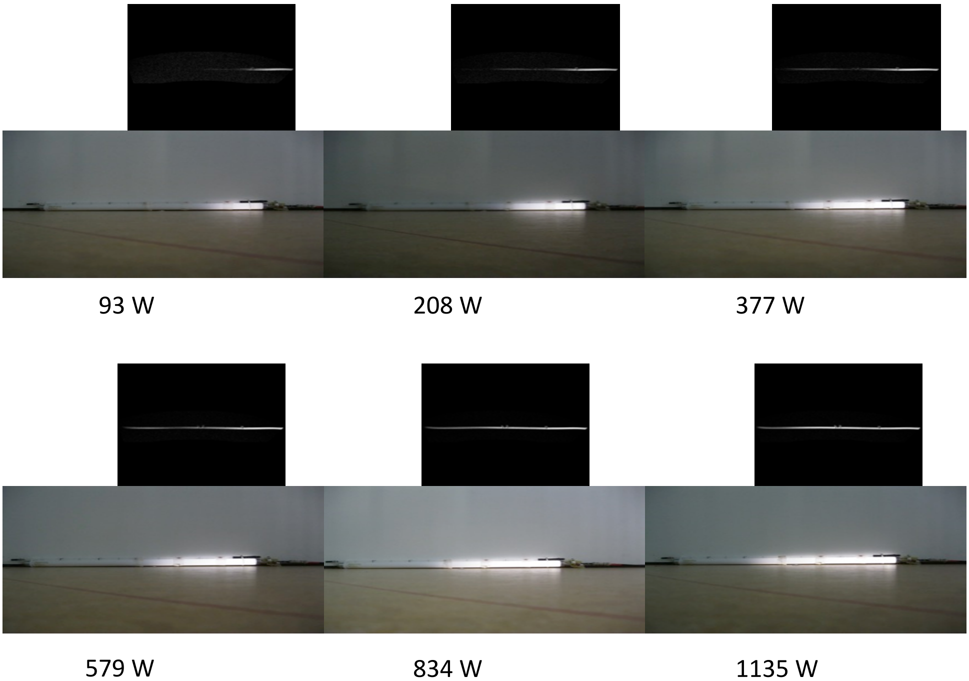

Figure 6.4: Demonstration of non-linear plasma based MRI coils (unpublished data). The plasma coil consists of a fluorescent tube that is connected to the RF transmit power of the MRI scanner via a 50 Ohm line and a surfatron. In order to visualize the extension of the B_1 field a small tube filled with water was attached coaxial to the fluorescent tube. Then different transmit pulses were sent with different power levels ranging from 93 to 1135 Watts. The glow was photographed, and the same experiment repeated with the same powers in the MRI scanner and a very large field of view to cover the entire length of the plasma coil. The length of the visible water phantom corresponds very well to the length of the plasma glow and is thus directly proportional to the applied power.

It is already known from Rayner et al. [34], that the length of a plasma column (argon plasma at 500 MHz) depends on the power applied. They were able to show that the length of the column increases as the square root of the applied RF power. They conclude that these findings will help to design an electronically controllable length that would allow for rapid reconfiguration for different frequencies.

6.2.6. PLASMA BASED RF SHIELDS

RF shielding is usually achieved using copper shields. When using such shields for shielding a coil, the number of eddy currents needs to be minimised to prevent the formation of image artefacts. One interesting property of plasma is that it can be transparent as well as opaque for certain RF frequencies. The plasma frequency can be reconfigured by changing, for example, the pressure inside the plasma coil. If an electromagnetic wave hits a plasma two outcomes are possible, which depend on the frequency of the incident electromagnetic wave and the plasma frequency [17, 18]:

1. If the frequency of the incident ω_i is larger than the plasma frequency ω_p the incident wave will pass through the plasma

2. If the frequency of the incident ω_i is smaller than the plasma frequency ω_p the incident wave will not pass through the plasma

One could use this property of plasmas to build switch-able RF shields like in figure 6.5. Such shields could be used to shield and/or decouple transmit/receive arrays. When the plasma is switched on it provides RF blocking like a conventional copper shield that helps to reduce radiation of power away from the element during transmit. During the receive, the switch can be switched off to improve receive sensitivity of the element. Such a switchable plasma based RF shield would address the problem of increasing radiation losses of RF coils at higher field strength MRI and also would help to protect body parts that are outside of the field of view from radiation. One could imagine, for example, switch-able shields incorporated into the body coil to reduce SAR in the arms and shoulders.

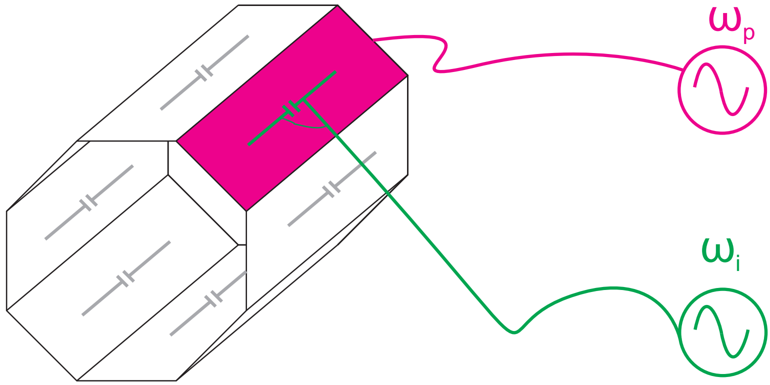

Figure 6.5: Principle of a plasma conductor based RF shielded coil. The copper conductor is removed and replaced by a plasma conductor sheet. The coil would consist of 8 transmit elements with 8 respective plasma shields. Each shield can be switched on individually.

6.2.7. RECONFIGURABLE PLASMA RF COILS

Plasma coils can be reconfigured during operation [1, 15, 16, 23, 29, 33]. One can for example change the gas pressure. By changing the gas pressurethe plasma frequency becomes shifted and thus the optimal noise frequency band. It is thus conceivable to use this feature to build dual tuned coils. One nice feature of a plasma antenna is that it can be switched off. In the off state the conductor completely vanishes, what remains is an space with a low pressure gas. This low-pressure gas has of course a much smaller radiation absorption coefficient than metal. In this way very transparent coils could be build. One could imagine using such a coil for example for PET-MRI or SPECT-MRI.

It might also be possible to drive one and the same plasma coil with two different RF frequencies: simply by using both ends of a plasma tube whereby at one end the surfatron for frequency one (maybe proton), and on the other end the surfatron for frequency two (maybe phosphor), could be excited.

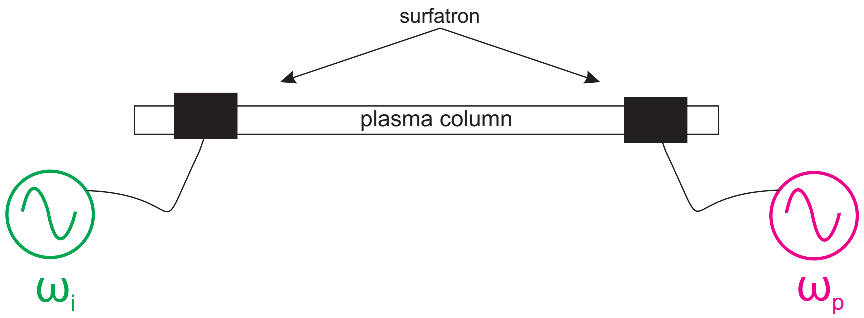

Figure 6.6: Principle of a double tuned plasma MRI coil: a single plasma column is fed be two different frequencies via a separate surfatron for each frequency.

This arrangement is interesting because it has been shown in the literature that a plasma antenna can be pumped. This means that the plasma is produced by a higher frequency RF source igniting the plasma while a lower frequency uses the ignited plasma column as an antenna.

6.2.8. Hybrid Plasma / Lumped Element RF Coils

It is also conceivable to combine established lumped element RF coil technology [24] with a plasma conductor. One could for example make the rungs of a birdcage resonator [8] from plasma conductors. This would then allow for perfect decoupling, For example for making a coil double tuned or for use with a receive array. Switching circuits would be no longer needed, and thus construction of transmit birdcages simplified.

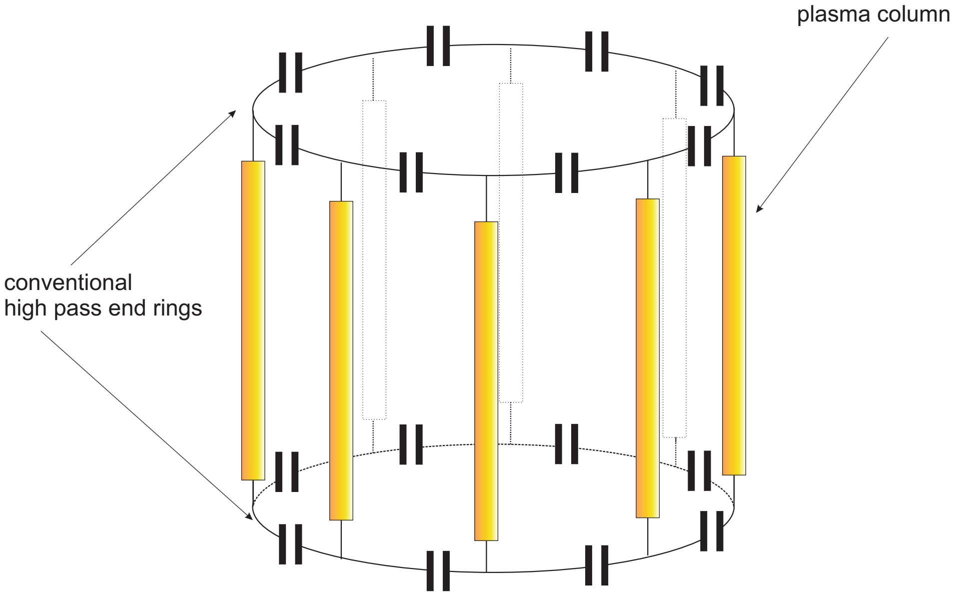

Figure 6.7: Principle of a hybrid plasma / lumped elements birdcage MRI coil: The single rungs of the birdcage are replaced by plasma columns. If the columns are switched on e.g. into the plasma state, the birdcage resonator will work like a conventional lumped elements birdcage. If the plasma in the columns is switched off, the birdcage modes disappear from the RF spectrum: only the modes of the end-rings are still present.

If a birdcage such as shown in figure 6.7 were to be realized it, could be effectively detuned to be used with a receive array when switching the plasma columns off.

6.2.9. DOUBLE TUNED BIRDCAGE COILS

Ifa birdcage as suggested in figure 6.7 were to be built, it could also be used in another interesting way: one could switch on only single legs. In this way, with the same capacitance in the end-rings but a different inductance, other modes of the birdcage resonator on other frequencies could be realized.

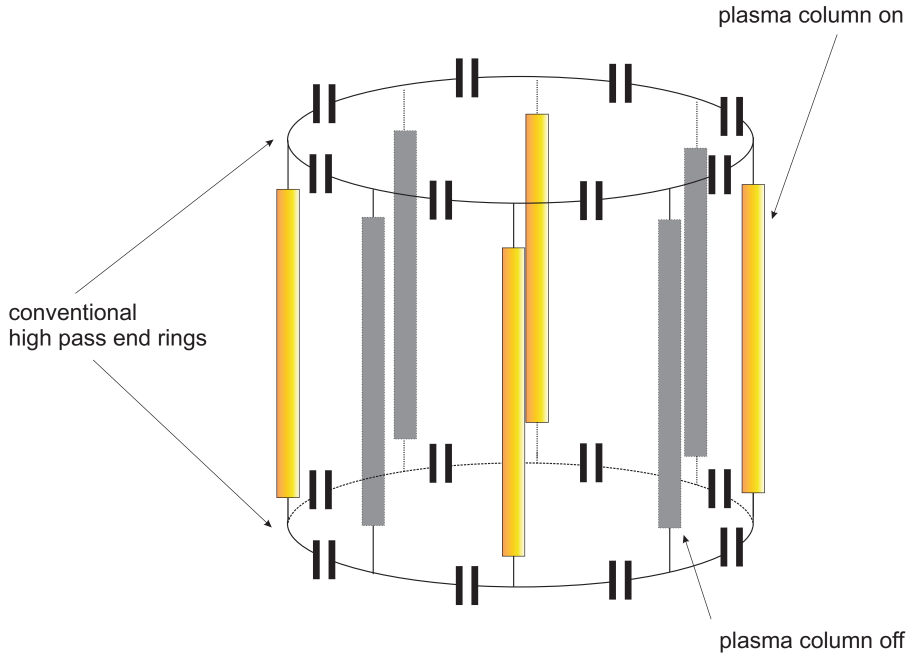

Figure 6.8: Principle of a double tuned hybrid plasma / lumped elements birdcage MRI coil: The single rungs of the birdcage are replaced by plasma columns. The columns can be switched on and off individually e.g. birdcage resonator will work as a four-rung birdcage at one time but can be also switched on to become an eight rung birdcage.

Such a birdcage could use modes that are associated with a small number of switched-on rungs for proton imaging, as there is sufficient signal, and all rungs for e.g. phosphor imaging. Again, such a solution would have the inherent beauty of not needing any detuning circuitry, when one wants to combine such a birdcage with a dedicated receive array.

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