Catheter Coils

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1 INTRODUCTION

Magnetic resonance imaging (MRI) is a very powerful tool for diagnostic imaging; however, it is very seldom used for guiding interventional procedures. On the other hand, in many procedures the quality of the guiding image modality is less than perfect, and therefore, MRI can be considered as an alternative.

Among these interventional procedures, the percutaneous cardiovascular procedures have a special place. Today, most of these procedures are conducted under the guidance of X-ray fluoroscopy. Using this guidance method, the catheters include guidewires that are visible, but this projection-based imaging modality provides very little information on the tissue or organ of interest. MRI appears to be a natural alternative to X-ray based imaging for this purpose because of its high tissue contrast. In addition, MRI has no ionizing radiation. High X-ray exposure to the patient, especially to children, can be harmful. This is also a significant problem for the physician who is conducting the procedure on a daily basis. MRI solves this problem as well. Therefore, MRI has a great potential to replace X-ray in guiding some of the percutaneous cardiovascular procedures.

While the advantages of MRI over X-ray fluoroscopy are apparent, it is also very obvious why currently MRI cannot replace X-rays. The patient access is poor in MRI scanners, although development of short and wide-bore magnets alleviated this problem. MRI-compatible equipment such as patient monitoring systems has limited availability. Most catheters and guidewires are not compatible with MRI. While these or similar problems are real, they are not fundamental limitations. The most important limitation of MRI compared with X-ray fluoroscopy is its poor visualization of the interventional devices. In MRI, most interventional devices appear dark, and therefore it is possible to visualize them only if a thin slice imaging method is used. The interventional devices get lost in the body when thick slice imaging is used.

This very fundamental problem of the poor visualization of catheters gave rise to the research field of catheter tracking under MRI. This effort resulted in the development of the catheter coils, i.e., the catheters that can be tracked using the MRI signals that are received by small coils embedded inside the catheters.

The development of catheter coils did not have a single motivation. The other motivation is to obtain high-resolution images of the blood vessels. As is well known in the field of MRI, the shape, size, and position of the receiver radiofrequency (RF) coils have critical importance for the signal-to-noise ratio (SNR) of the acquired images. With rather straightforward manipulation of the pulse sequences, the increase in the SNR can be used for higher image resolution. As the RF coils get closer to the point of interest, the SNR is expected to be improved and therefore higher image resolution can be obtained. With this motivation, researchers have investigated the placement of small receiver coils, so-called catheter coils, inside the blood vessels.

The interest in development of catheter coils developed immediately after commercial MRI scanners became available. In 1984, Dr Howard Kantor, while working at the NIH, tested the first catheter coil concept for increasing the SNR in 31P NMR spectra.\(^1\) As is well known, the 31P signal is very weak compared to the signal of proton and therefore obtaining a clinically useful result is often difficult. On the other hand, one of the key components in the SNR in a magnetic resonance experiment is the position and the design of the receiver coil. SNR can be increased significantly by reducing the size of the coil and placing it close to the region of interest. In Dr Kantor’s case, the region of interest was the heart. He and his colleagues placed a small loop (7.5 × 24 mm, two turns) inside the heart. They tuned the coil using a single capacitor and obtained 31P spectra of the myocardium, as shown in Figure 1. This pioneering approach triggered interest in catheter coils and resulted in many more publications on the subject.

Three research groups led by Dr Gregory C. Hurst of Case Western Reserve University,\(^2\) Dr Alastair J. Martin of the University of Toronto,\(^3\) and Dr Krishna Kandarpa of Brigham and Women’s Hospital\(^4\) demonstrated the use of catheter coils for high-resolution imaging of blood vessels almost simultaneously. While the designs of Dr Hurst and Dr Martin were based on opposed solenoid coils (Figure 2), the design of Dr Kandarpa was a small rectangular loop (elongated loop) placed in an 8 Fr (1 Fr is 1/3 mm) catheter (Figure 3). Using this design, Dr Kandarpa showed high-resolution images of excised human arteries. These designs are discussed later.

In the mean time, the first catheter tracking was initiated by Dr Jerome L. Ackerman of Massachusetts General Hospital (MGH) with his pioneering work published as a 1986 Society of Magnetic Resonance in Medicine Annual Meeting abstract.\(^5\) In this work, Dr Ackerman explains the use of a small RF coil for tracking the position of interventional devices (Figure 4). Later, this work was extended by Dr Charles L. Dumoulin in a 1993 article in an elegant way.\(^6\) In this method, Dr Dumoulin applies non-slice-selective RF pulses followed by three orthogonal readout gradients to obtain the projection of the catheter in three orthogonal axes. From this information, it becomes rather trivial to find the location of the tip of the catheter. This method was later used by many researchers.

In the next section, the designs of catheter coils are discussed.

2 CATHETER COIL DESIGNS

Researchers have developed many catheter coil designs. In the evaluation of the performance of these designs, in
addition to their ability to help track the position of the catheter and/or acquire high-resolution images of the blood vessels, their mechanical properties are of critical importance. We start this section with a discussion on the mechanical properties of catheters.

One of the most important mechanical properties of catheters and guidewires is their dimension. The dimension of a catheter is usually measured in Fr (abbreviation of French), which determines its maximum diameter (a 3 Fr catheter’s diameter is 1 mm). While a 15 Fr catheter can be considered very large, a 3 Fr catheter is small. Size selection depends on the specific application.

Most catheters have one or more lumens for different functions. Some lumens are used for holding guidewires and others for injecting therapeutic or contrast agents. These lumens can also be used for delivering other vascular devices to remote body parts. Obviously, the dimensions of these lumens are also critical. For example, a standard guidewire diameter (usually measured in inches; 0.035, 0.025, and 0.014 in. are typical guidewire dimensions but other dimensions are also used) needs to be used for the lumens that hold guidewires.

Stiffness is a critically important mechanical property of catheters, and it varies greatly with the application of the device. A typical guidewire has a stiff shaft for improved torque (ability to turn the distal end of the guidewire by rotating its proximal end) and pushability (ability to transfer the longitudinal force from the proximal end to the distal end); however, the distal end of the guidewire should be very soft for entering into delicate vessels without damaging them. Guiding catheters have similar stiffness properties. These properties of guiding catheters and guidewires enable the interventionists to navigate inside the body. Other types of catheters, such as balloon catheters, are navigated inside the blood vessels with the help of guidewires and guiding catheters, and therefore their torque or pushability are not critical, but they should be as soft as possible.

The above-mentioned mechanical properties determine the kinds of RF coils that can be embedded inside a catheter. Obviously, a large and rigid RF coil is not desirable inside a flexible and thin catheter, since it may alter the mechanical properties of the catheter. In the next subsections, various designs that can be placed inside catheters are discussed. We will start with the design of the elongated loop.
2.1 Elongated Loop Design

As mentioned above, flexibility is of critical importance in catheter coil design. However, when a coil flexes, its inductance may change, and therefore the tuning of the coil and its performance may degrade. In order to resolve this issue, a conductor loop is formed using two parallel wires by shorting them out at the distal end and using the proximal end as the terminal of the loop coil. Since the inductance of this design does not change significantly when the wires flex, it is ideal for a catheter coil design.

The early demonstration of this design by Atalar et al.\textsuperscript{7} is shown in Figure 5. In this design, the two capacitors, which were used for tuning and matching of the coil, respectively, were placed right at the terminal of the coil. While the coil itself was 4 Fr in size (diameter: 1.33 mm), the maximum diameter of the design (around the region where the capacitors are placed) was 9 Fr (diameter: 3 mm). Decoupling is achieved by placing a shunt PIN diode on the coaxial cable. The position of the diode is adjusted to minimize induced currents on the leads when the PIN diode is on. The design is further miniaturized in Ref. 8 to 5 Fr by using microfabrication techniques. Both in vivo and ex vivo tests were conducted to demonstrate the performance of the design.

In order to understand the sensitivity of a coil, one can use the reciprocity principle and assume an imaginary unit current applied at the terminal of the coil and use the transverse component of the magnetic field created by this current as the sensitivity of the object. When the magnetic field is normalized by the square root of the real part of the impedance measured at the terminal of the coil, the sensitivity of the coil can be obtained.

The sensitivity of elongated loop designs as RF receiver coils depends on many parameters. This design provides the best quality images when the axis of the catheter is along the main magnetic field. In this case, the sensitivity of the catheter drops with the square of the distance to the catheter; it is relatively uniform along its length and also in the circumferential direction. When the orientation...
of the catheter is not along the main magnetic direction, circumferential sensitivity variations are observed. In addition, when the length of the coil increases, sensitivity variation along its length can be observed. In the designs reported in the literature, approximately 10 cm long catheter coils are used in 1.5 T MRI scanners. The separation between wires plays a critical role in the coil’s performance. An RF coil designer who designs surface coils knows that the diameter of a coil should be approximately the same as the depth of the point of interest in the body. One may form an analogy between the wire separation of an elongated coil by the surface coil’s diameter, but in this case since the wire separation is always smaller than the point of interest (typically wire separation is 0.5 mm whereas the point of interest is typically farther away than 5 mm), the SNR of the acquired images with elongated loop designs increases with increase in the wire separation. In addition to these parameters, the dielectric material that covers the coil has a significant role in the performance of the design. As might be expected, the performance of the design improves with thick insulation. The insulation material, especially around the tuning and matching capacitors, should have low losses. Teflon is a preferred material for this purpose. The wire diameter also is critical in the design. The losses in the coil are mainly due to the resistance of the wire (eddy current losses are typically negligible in this design), and therefore thick wires are preferred. Because of its MR compatibility and superelasticity properties, one may consider nitinol as a conductor material but its rather poor conductivity makes it a nonideal choice. Since the skin depth at the Larmor frequency is only a few micrometers, one may consider using silver- or gold-plated nitinol for this purpose. On the basis of the mechanical design constraints, the designer needs to optimize the wire length, diameter, separation, and insulation thickness. As mentioned above, proper choice of the wire and insulation materials is also important.

When manufacturing an elongated loop coil, fixed-value capacitors are typically used for tuning and matching, mainly because of their small size. The desired capacitance values can be obtained by placing multiple capacitors in parallel. One alternative method is to adjust the length of the coil for a given tuning capacitance value. As in the surface coil tuning process, distributing the capacitance around the elongated loop is a preferred method. This method not only reduces the load dependence of the resonance frequency but also improves SNR, although only slightly because in the elongated loop design the main loss source is the conductivity of the wires. Since the quality factor of an elongated loop design is rather low (around 10), tuning and matching is relatively easy compared to that in surface coils.

When calculating the sensitivity of the coil, unbalanced currents should be considered. Unbalanced currents are the currents on the outer surface of the coaxial cable. If a current is applied to a transmission line with two parallel wires, it is expected that the currents on the wires are equal in magnitude but opposite in direction. In this condition, the current is considered to be “balanced” and all the electrons flowing from one wire return from the other one. If a coaxial cable is used, when equal and opposite currents flow on the inner conductor and the shield, the magnetic field generated by this balanced current is totally trapped in the dielectric material of the coaxial cable and no magnetic field can be observed outside the shield. On the other hand, if the currents on the inner conductor and the shield are unbalanced, the excess current flows on the outer surface of the shield. The magnetic field generated by the current in the wire is not totally trapped inside the dielectric of the coaxial cable but leaks into the medium around the coaxial cable. When a coaxial cable is connected to an electrically asymmetric elongated loop coil, unbalanced currents are observed that reduce the SNR of the acquired images and cause asymmetrical sensitivity in the circumferential direction. In order to eliminate these unbalanced currents, electrical symmetry needs to be maintained, and also, when possible, balun circuits should be used. Since balun circuits are typically very bulky, a bazooka balun using double-shielded coaxial cables can be used. As can be understood from the above discussion, dealing with unbalanced currents is often difficult. If it is ignored, they cause significant image artifacts and SNR degradation.

Tuning and matching of the elongated loop design is a significant problem. If one wants to achieve the maximum SNR, the tuning and matching capacitors should be placed at the end of the loop before connecting the elongated loop to the coaxial cable. To reduce the unbalanced currents, a symmetrical tuning/matching circuit is preferred. Since the presence of this circuit reduces the flexibility and increases the diameter of the design, alternative approaches are discussed in the literature. Placing a single tuning capacitor without a matching capacitor inside the catheter is an alternative if the designer accepts an associated SNR loss. Some designers place no capacitors in the catheter and tune the design remotely by placing the related circuit elements in a box outside the catheter. This results in a significant performance loss, but the flexibility of the design is not compromised. In cases where the inductance of the loop changes when the catheter flexes inside the body, adjustment of tuning may be necessary. Electronic tuning using a varactor diode has been proposed to solve this issue. Although this is an attractive alternative to fixed capacitors, the varactor diodes generate additional noise and degrade the performance of the design. The resulting compromise must be considered during the design. In short, in tuning an elongated loop, the designer should consider all the mechanical and electrical constraints.

One other important aspect of the elongated loop design is decoupling the design from the transmitter coil (this is typically a birdcage coil). As mentioned earlier, PIN diode-based decoupling circuits are used in the elongated loop designs: however, this alone is not enough to obtain the desired decoupling. Induced currents on the coaxial cable of the design should be handled with care. As mentioned when describing the sensitivity of the coil, these unwanted induced currents may be decreased using balun circuits. If there is strong coupling between the transmit coil and the elongated loop design, it may cause significant amplification of the RF electric field around the coil, leading to excessive heating and possibly a burn. Testing of the design in adverse conditions is necessary to ensure patient safety.

In the literature, elongated loop design can be seen in many forms. One of the most interesting designs is the expandable coil. As was previously mentioned, as the separation between two parallel wires increases, the SNR of the acquired images increases. On the other hand, the size of the catheter limits the maximum wire separation in the elongated loop design.
2.2 Loopless Design

Miniaturization of catheter coils is a challenge. As discussed in the elongated loop design, reducing the catheter size reduces the SNR performance of the design, since the separation between the wires decreases. In order to solve this problem, the loopless design has been proposed. In this design, the current is applied to an antenna that does not have a closed loop. One may think that the loop is closed by conduction through the body. Since there is no loop in the design, the common name of “coil” cannot be used for this type of RF receivers, but instead the word “antenna” is used.

The loopless antenna design was first described by Ocali and Atalar as a coaxial cable with an extended inner conductor. Since no other elements are involved in this design, the diameter of the antenna is solely determined by that of the coaxial cable. Although very thin coaxial cables exist, mechanical constraints limit the minimum diameter of the design. In Ref. 13, a 0.014 in. diameter loopless design was reported in the form of a guidewire.

In Ref. 12, the SNR performance of the loopless design was optimized by adjusting the whip length (extending the inner conductor beyond the end of coaxial cable). When a bare wire is used, the optimum whip length becomes approximately equal to a quarter wavelength where the wavelength is measured in the body at the Larmor frequency. This value is approximately 10 cm for 1.5 T and 5 cm for 3 T scanners. As described by Susil et al., the optimum whip length can be decreased by coiling the whip and increased by insulating it with a dielectric material. Note that the dielectric constant of the body (and water) is typically much higher than that of any dielectric material one may use. Even a very thin layer of insulation has a very strong effect on the optimum whip length.

As in the elongated loop design, the loopless design outperforms the loopless design when imaging regions in proportion to the distance to the antenna. Note that, in the elongated loop design, the sensitivity drops approximately equal to the square of the distance. Therefore, while the elongated loop design outperforms the loopless design when imaging regions close to the catheter, its performance degrades rather quickly, and further away from the catheter the loopless design outperforms the loopless design. The sensitivities of typical designs are approximately equal when the point of interest is at a distance of 7 times the wire separation of the loop design.

The sensitivity of the loopless design along its length shows a significant variation. At the junction point (where the whip of the antenna starts and the coaxial cable ends), the sensitivity is maximal, whereas at the tip of the whip the sensitivity is nearly zero. The sensitivity also drops toward the proximal end of the loopless antenna. When the shaft (coaxial cable) is insulated, the sensitivity is extended. This property of the loopless antenna has been explained by Susil et al.

One of the main problems with the loopless antenna design is that its sensitivity is very low at its distal end. To alleviate this problem, Qian et al. showed that tapered insulation for the whip could be used. Placing a short solenoid with a single connection to the end of the whip without a return path has also been proposed by Serfaty et al. for using this design as a guiding catheter.

An important property of the loopless antenna is that its optimum impedance is close to that of a typical coaxial cable, and therefore with remote tuning and matching no significant loss of performance is observed. In fact, in many instances it can be directly connected to the preamplifier. As in the elongated loop design, a PIN diode can be used for decoupling purposes. The decoupling circuit helps in reducing the current induced on the loopless antenna during transmission of RF pulses to the body coil. A PIN diode placed as a shunt between the inner conductor and the shield of the coaxial cable a quarter

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This article is © 2011 John Wiley & Sons, Ltd.
This article was published in the Encyclopedia of Magnetic Resonance in 2011 by John Wiley & Sons, Ltd.
DOI: 10.1002/9780470034590.emrstm1135
The opposed solenoid is the third basic type of catheter coil. By placing two solenoids facing each other, a relatively uniform field is obtained outside the catheter.3,7,18 This idea was termed also as “inside-out coil” to emphasize this property. An early use of this design was for the investigation of oil wells.19

A most interesting property of this design is that, when it is aligned with the main magnetic field, it provides a fairly uniform sensitivity in the region between the solenoids. This improved uniformity comes at the cost of SNR performance degradation. In the publication by Hurst et al.2 (Figure 7), a field-effect transistor was placed inside the catheter after a tuning capacitance. Placing a preamplifier as close as possible to the catheter not only decreases the cable losses but also, more importantly, decreases the sensitivity of the design to inductance changes.

All other properties of this design are very similar to those of the elongated loop design.
This idea was later extended for possible visualization of stents.\textsuperscript{22} Since stents are usually made out of metals, with a slight change in their structure and by adding a tuning capacitor they may become visible under MRI.

One issue related to ICRF coils is that active decoupling circuits cannot be used with this design since there is no wired connection to the scanner hardware. Although in some cases coupling during transmission may be used as an advantage, it may also be a safety problem, and also a high flip angle around the catheter may cause image artifacts. This problem can be solved using passive decoupling, e.g., by using back-to-back Shottky diodes or low turn-on voltage fast switching diodes.

With this technique, the tissue around the ICRF coil appears bright compared to the rest of an image acquired using a surface coil. When the MRI signal is acquired using multiple surface coils, it is possible to extract the ICRF-related signal from the rest of the signal. This is best described as using reverse polarized reception. In MRI, the echo signal comes circularly polarized. If a receiver is tuned to be sensitive to reverse circular polarization, no signal is expected to be received except when there is an ICRF coil inside the body. The ICRF coil picks up the forward circularly polarized signal and creates a linearly polarized reactive field. It is known that a linearly polarized field can be thought of as the sum of two opposing circularly polarized fields (forward and reverse). Since the receiver coil is tuned to the reverse field, it does not detect any signal directly originating from the body but it detects the signal originating from the ICRF coil. When multiple receiver coils are used, signals acquired from these coils can be combined such that only the reverse sensitivity remains and therefore only the signal from ICRF coil remains on the images. A detailed description of this complex phenomenon can be found in Ref. 23, but, in short, if one acquires the ICRF coil images using multiple coils, the background information can be separated from the ICRF images by an image processing technique. This is especially important if one wants to track the position of a catheter inside the body accurately.

3 APPLICATIONS

Currently, the catheter coils are mostly used as parts of investigational devices. There are, however, many applications that may utilize catheter coils.

3.1 Imaging

One of the basic problems of catheter coils is improving the image quality and resolution by increasing the SNR. Comparing the above designs is not possible without knowing the exact conditions in which each design will be used. For example, if space is very limited, the loopless design should be considered. One should also always examine whether the same or similar image quality could be obtained as when using surface coils.

With the aim of quantifying the SNR performance of internal coils with respect to the best surface coil, the ultimate theoretical value that one can obtain using a surface coil has been calculated in a study by Ocali and Atalar.\textsuperscript{24} It is known that the MRI signal is related to the transverse component of the magnetic field generated by a coil when a unit current is applied at the terminal of the coil, and the noise is related with the electric field generated by the same coil under the same conditions. To find the maximum SNR value, the magnetic field needs to be maximized and the electric field needs to be minimized. In their calculation, an optimization problem was solved by assuming that there is an infinite freedom of designing surface coils. In solving this problem, instead of trying to find the optimum coil geometry, only the electromagnetic field inside the body was considered. In the Ocali and Atalar article,\textsuperscript{24} the ultimate value of the intrinsic SNR was solved numerically and various existing coil designs were tested under different conditions. The study showed that the current technology is not far from the maximum attainable SNR. In the article by Kopanoglu \textit{et al.},\textsuperscript{25} this study was extended to obtain an analytical formula for the ultimate intrinsic SNR. As can be seen from these formulations, the maximum attainable SNR reaches infinity at the surface of the body and decreases proportional to the 2.5th power of the...
distance from the surface. To achieve an SNR value higher than this value, an internal coil (or catheter coil) is necessary. Similar to this ultimate intrinsic SNR calculation, a calculation for the internal coils can be conducted by assuming that there is a cylindrical volume at the center of the body in which one can place the internal coil. In Figure 9, a comparison of the maximum attainable SNR using only surface coils with the SNR produced by a combination of internal and surface coils is shown. On the same graph, the performance of a loopless design is also shown, to demonstrate its relative performance compared to the maximum attainable SNR. As can be seen from the graph, the loopless antenna outperforms even the best surface coil when the point of interest is close to the center. However, it seems that the loopless antenna is far from the best internal antenna. The other designs have to be investigated, but placing an antenna in a 0.75 mm diameter hole (as in the example shown in Figure 9) is a challenge.

One of the main problems with internal MR coils is that their sensitivities are not uniform and the images obtained using these coils cannot be used for diagnostic purposes unless the signal intensity variations are corrected. In some publications, signal intensity correction techniques are used to show high-quality images; however, currently all automated signal intensity correction methods also degrade the underlying images and therefore are not preferred.

3.2 Tracking

One of the most important reasons to embed a catheter coil in a catheter is to visualize the catheter’s position during MRI. Real-time tip-tracking techniques have been implemented on commercial scanners for finding the position of small solenoidal catheter coils. The general method that was proposed by Ackerman et al. is still valid, although some novel improvements have been developed. Embedding coils inside the catheter for the purpose of finding the position of a catheter is termed active catheter tracking. It should be noted that “passive catheter tracking” is an alternative to active catheter tracking. In the passive tracking technique, a marker that may cause an image artifact is positioned on the catheter. Although this technique is relatively simple to implement, finding the position of the artifact is not trivial during a complex vascular procedure. If the artifact is large, it obscures the underlying image and therefore is not preferred, but if it is too small, it becomes hard to visualize and causes loss of the catheter in the body. Active catheter tracking solves this issue at the cost of increased complexity and RF safety problems, which will be discussed later.

In addition to the solenoid, the loopless antenna design has also been used for active tracking. This method enables complete visualization of the catheter shaft with a high frame rate. Later, this method was extended to include visualization of the catheter within its background while updating the catheter images in a high frame rate, with background information acquired at a lower rate. Guttmann et al. used active catheter tracking within a three-dimensional visualization tool.

Active catheter tracking techniques have been proposed for different catheter designs. Guiding catheters and guidewires became MR-visible with the help of the loopless design, and this pair was used to guide a balloon angioplasty procedure. Again, this design was used inside a vascular needle to guide septal puncture and also for creating a shunt between the portal vein and vena cava. An injection catheter has been manufactured using a similar design and used to inject therapeutic agents into the heart. Since this article is more geared toward the technical aspects of catheter coils, these specific applications are not discussed in detail.

4 SAFETY ISSUES

The use of catheter coils inside the human body raises safety issues. In testing the effectiveness of the design, it is always necessary to check whether there are any safety problems. When verifying the safety of a catheter coil design, it is important to understand the issues related to RF heating.

It is very well documented that, when a bare wire with a resonant length is placed inside the body during an MRI examination, a significant temperature rise at the tip of the wire can be observed. Even when these long metallic objects are insulated, some significant heating may be observed. As discussed earlier, induced currents on the catheter coils can be prevented by using active and passive decoupling circuits as well as balun circuits. In order to decrease the possibility of excessive heating, alternative methods have been proposed including optical transmission lines. Extensive tests in worst case conditions should be conducted on phantoms before using these designs on humans. In a typical phantom safety test procedure, the highest allowed specific absorption rate (SAR) is applied to the phantom while the catheter coil is inside the phantom and the temperature rise in the immediate vicinity of the catheter coil should then be measured. Note that the SAR applied to the phantom should be measured rather than trusting the manufacturer-supplied values since...
the latter are often higher than what is actually applied to the phantom. It should also be noted that the position of the catheter coil inside the body may vary and therefore it is advisable to conduct these tests in all the extreme positions that the catheter coil may take up in the body. When the temperature rise is less than 2 °C over an extended period (20 min), one may assume that catheter coil is safe to use.

Gradient-induced currents on catheter coils may cause peripheral nerve stimulation. Although it has not been documented yet, it is feared that these currents may be at an important concern.

While embedding a coil inside the catheter, the mechanical safety of the design needs to be investigated. Especially when embedding electronic circuit elements such as capacitors and diodes inside a catheter, the mechanical properties of the catheter may be compromised. Mechanical tests well known to catheter engineers should be applied to these designs. These tests include fatigue, integrity, torque, and pushability.

When these problems are carefully investigated, catheter coils can safely be used in diagnostic and interventional procedures.

5 CONCLUSION

In this article, a summary of the current catheter coil designs was given. Some technical details are omitted in order to make the text readable. A reader interested in the design or use catheter coils should also read original articles on this subject, some of which can be found in the bibliography of this article. In addition to the basic catheter coil types that were discussed above, the reader may find other designs. The content of this article should be able to help the reader in understanding the other catheter coil types that are not mentioned here.

6 RELATED ARTICLES

Bioeffects and Safety of Radiofrequency Electromagnetic Fields; Coils for Insertion into the Human Body; Design and Use of Internal Receiver Coils for Magnetic Resonance Imaging; Jolesz, Ferenc A.: The Development of Interventional MRI; RF Device Safety and Compatibility

7 REFERENCES

Biographical Sketch

Ergin Atalar. b. 1961. BS, 1985, MS, 1987, PhD, 1991, Ankara, Turkey. Immediately after graduation from Bilkent University, joined the Johns Hopkins University, where he became a Professor of Radiology, Biomedical Engineering, and Electrical and Computer Engineering, and Director of the Center for Image Guided Interventions. Currently, a Professor in the Department of Electrical and Electronics Engineering and the Director of National Magnetic Resonance Research Center, Bilkent University. Also a founder of MRI Interventions, Inc. and Troyka Med AS. Main research interests include magnetic resonance imaging and image guided interventions.