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MAGNETIC FIELDS IN SURGERY*

S. J. St. Lorant
Stanford Linear Accelerator Center, Stanford University
P. O. Box 4349, Stanford, California 94305
U. S. A.

SUMMARY

Magnetic fields generated by conventional and superconducting magnets have been used successfully in numerous surgical procedures. These applications include magnetically guided catheters for use in the cerebral vascular system, permanent implantation of magnetic materials for purposes of atrial stimulation and sensing, and the delivery and temporary retention of magnetic emboli. This last technique has received considerable attention since it was found that suspensions of magnetic materials in vulcanizable surgical silicone not only can be used to obliterate cerebral malformations but have many potential applications in tumor therapy. The magnetic systems which have been developed for these purposes are described.

INTRODUCTION

For the purposes of this report it is convenient to classify the various applications of magnetic fields in the medical arts according to the qualitative restrictions imposed on the type of magnetic field, and thus on the means of producing it.

Specifically, we distinguish three separate areas of use. Firstly, we have energy and signal devices which usually involve some form of inductive transformer to provide the necessary transcutaneous magnetic coupling. Typical applications include contactless nerve stimulation and signal detection, high power electromagnetic energy feed systems for a totally implanted device - principally artificial hearts and pacemakers, and the electromagnetic measurement of the flow of blood and other body fluids.

Secondly, we recognize magnetic systems whose purpose it is to provide controlled guidance and propulsion of catheters into difficult to reach parts of the body. Examples of this use are the magnetically guided catheters which have been used to gain passage into the cerebral circulation, to provide rapid and safe access to selected bronchial segments, and even to give the necessary positive contact in endocardial pacing.

Thirdly, we have systems in which the magnetic field fulfills a retentive function: ferromagnetic material is held in a given position in the vascular system until a certain biological process has taken place. The occlusion of intercranial aneurysms by ferromagnetic thrombi and the ferrosilicone occlusion - techniques now finding application in tumor therapy-are representative of this application.

The first class is of limited interest to the magnet designer. The emphasis is on the inductive coupling over a wide range of frequencies, and the quality of the magnetic field generated by the device is of little importance. Therefore we will not discuss this application further.

In the other two classes, the magnetic field, its generation and use in conjunction with conventional surgical equipment, is of basic importance. Because guidance systems exploit the properties of magnetic field gradients, while retentive systems rely more on uniform fields, we discuss them separately, even though the forces exerted by the field on the ferromagnetic body in both cases are described by the same basic magnetostatic equations. In guidance systems the forces tend to be limited by the achievable field gradients, whereas retentive applications usually require fields sufficiently high in absolute value to attain local saturation of the average magnetic moment of the injected material.

For completeness we might mention that the transport of particle beams in various devices used in clinical radiotherapy generally requires magnetic fields. However, as the magnets in this application do not directly interact with the human body, they do not fall within our present sphere of interest.

GUIDANCE SYSTEMS

The basic elements of a magnetic catheter are essentially the same for all applications although the dimensions and materials may differ. A catheter consists of a suitable length of relatively stiff polyethylene tubing to which is attached a short length of very flexible silicone rubber tubing. A small cylindrical permanent magnet with an axial perforation is inserted in the distal tip. A standard tubing adapter is attached to the proximal end to facilitate connection to various clinical hardware. The catheter is introduced into the body and guided to the point of application by the external field. Directivity is achieved by the alignment of the magnetic dipole (cylindrical magnet) in the external field, while the propulsion force results from a superposition of the forces due to the environment (e. g. , the flowing blood in the venous system, the moving air in the bronchial system, or even friction) and those due to the applied magnetic field. These forces can act separately; more usually they interact. Thus an alternating magnetic field will cause the magnetic tip of the catheter to oscillate. These oscillations will promote lateral motion of the flexible part of the catheter, which in the vascular system for example will combine with the blood flow to develop a propulsive force aiding the forward motion of the catheter. Figure 1 shows such a catheter introduced into the kidney of a dog through the renal artery.

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Fig. 1--Magnetic catheter in the kidney of a dog.

Experimental work has shown [1,2] that field gradients in excess of 4 Tm^{-1} are necessary for reliable manipulation of the catheter tips: such a gradient is at the limit of that which can conveniently be achieved by a moderately sized conventional iron-core solenoid. Montgomery and Weggel[3] have estimated that a solenoid 0.3 m in diameter, with a central field of 2 T, capable of producing a field of 0.7 T at an axial distance of 0.15 m from the magnet, and thus a gradient of some 7 Tm^{-1}, would consume about 800 kW of power. On the other hand, a superconducting coil with these characteristics is relatively easy to build: Hale et al.[4] have recently described in detail a 2T superconducting solenoid expressly designed for catheter guidance, and Kolm has discussed its performance at this conference. The Massachusetts Institute of Technology - Massachusetts General Hospital group has had much previous experience in the field: they have experimented with permanent magnets and conventional electromagnets, the largest of which was a 5 kW water-cooled magnet, generating a field gradient of 0.9 Tm^{-1}. Lately they have concluded[2] that the initial concept of "magnetic propulsion"[5] is no longer valid because the kinetics of catheter motion are primarily determined by the venous flow, with the tip magnetically deflected at critical junctions and magnetically fixed at its destination.

A rather different approach has been adopted by the Brookhaven National Laboratory magnet group[6] in the construction of a guidance magnet for the Neurological Institute of the Columbia Presbyterian Medical Center[7] in New York. This superconducting magnet can be rotated at speeds varying between 60 and 900 revolutions per minute, thereby providing the net driving force to the catheter. The magnitude of the field remains sensibly constant but its direction changes. This effectively results in a traveling wave field which acts on the catheter tip, causing it to flutter and to move forward with a swimming motion. The vibratory motion further limits the contact of the catheter with the vessel walls, thereby greatly reducing the frictional forces opposing the motion.

The rotating element of the system is a solenoid, about 0.1 m long, 0.17 m inside diameter, and 0.188 m outside diameter, wound with 25,429 turns of 200 μm diameter niobium-titanium superconductor. The wire contains 402 filaments, 7 μm in diameter, twisted at the rate of 85 per meter, and embedded in a copper matrix to give an overall copper-to-superconductor ratio of unity. The entire winding is surrounded by a wrap of fiberglass-epoxy, and housed in an iron shield which provides the flux return path. At 17.5 A, the peak field of the coil is 3 T, which translates into a useful field of 0.1 T at an axial distance of 0.23 m from the center of the coil. Figure 2 shows the magnet system, with the variable speed drive motor in the foreground. An interesting refinement is the persistent switch, which eliminates the inevitable cable tangle. The cooldown, energizing, and preliminary checks are done away from the angiographic facility, the magnet being rolled into the treatment center when required. Enough helium is stored in the dewar to permit 60 minutes of operation between fillings.

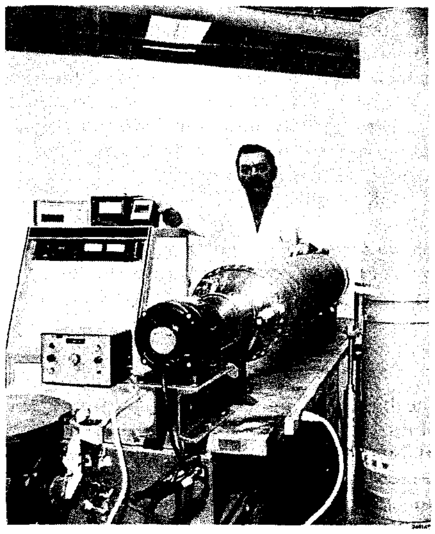

Fig. 2--The BNL rotating superconducting magnet for magnetic catheter guidance.

RETENTIVE SYSTEMS

The operative technique employs one or more magnets to hold a bolus of ferromagnetic material in place in a specific part of the venous system until local thrombosis is complete. The concept originated with the development of methods for the stereotactic thrombosis of intercranial aneurysms:[8,9] permanent magnets implanted in the vicinity of the lesions provided the necessary retentive forces. This approach suffered from many disadvantages, the principal two being the hazard to the patient during the open surgical procedure and the fixedness of the magnetic field. The need for an external, adjustable source of magnetic field thus soon became apparent.

The strength, size, and, hence, the weight of such a magnet are dictated by the field required at the aneurysm, assumed to be located at some distance from the magnet pole piece. Maximum depths typically encountered are around 0.1 m, for which flux densities of about 0.15 T are required at the injection point to prevent migration of the ferromagnetic material into the venous system. With the necessary safety factors this corresponds to about 0.2 T at 0.1 m from the pole piece. Clearly magnets with this kind of capability must be superconducting: a conventional electromagnet is just not feasible. Not only do the body geometry and surgical access preclude any iron return flux path yokes, but the sheer physical size of the equipment required to produce central fields in excess of 5 T makes its use in a typical operating room totally impractical.

Figure 3 shows the cross section of a superconducting magnet built for this application at the Stanford Linear Accelerator Center. The solenoid consists of seven double pancakes with a constant outside diameter of 0.115 m and a stepped inside diameter which increases from 0.015 to 0.055 m in six equal increments. The center of the coil contains an iron core similarly stepped, and grooved to provide passages for helium circulation. The material used is 5000 μm wide Nb_3Sn tape, stabilized with 50 μm of copper, and formvar insulated on the faces. The pancakes are edge-cooled only, and are separated from each other by 0.001 m thick radial spacers. At the design current of 210A, the field in the center of the face of the iron core is 7.3T. Table I illustrates the field distribution when the magnet is energized to 200A. Note that the center of the coordinate system is located on axis and 0.027 m away from the center of the face of the iron core, so that the above field criteria are satisfied.

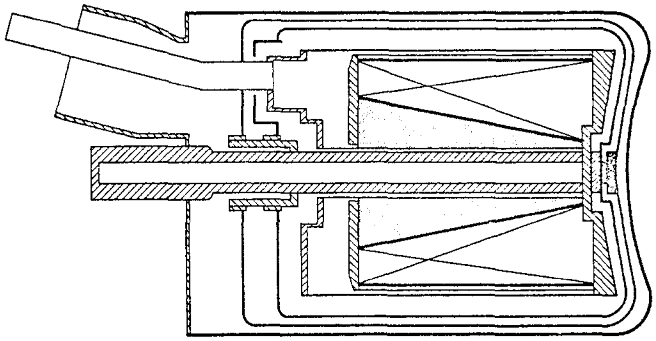

Fig. 3--Section through the SLAC Nb_3Sn superconducting magnet.

TABLE I

FIELD DISTRIBUTION AT 200A (T)

Axial Distance, cm	Radial Distance, cm 0	2	4	6	8	10
0	0.930	0.72	0.53	0.45	0.314	0.214
2	0.460	0.430	0.38	0.31	0.230	0.168
4	0.290	0.28	0.25	0.21	0.168	0.133
6	0.230	0.184	0.168	0.148	0.124	0.102
8	0.180	0.130	0.130	0.115	0.096	0.800
10	0.132	0.094	0.090	0.083	0.064	0.600

The magnet is housed in a 0.12 m diameter and 0.145 m long cylindrical helium vessel which in turn is surrounded by two copper radiation shields. This assembly is mounted concentrically in a 0.15 m diameter and 0.22 m long vacuum can. A vacuum-insulated flexible umbilical line joins this unit to a 25 ℓ capacity liquid helium storage tank which also carries the current leads and the transducers necessary to operate the magnet, such as magnet voltage sensing leads, resistance thermometers, liquid level gauges and level alarms, and a flowmeter to monitor the helium boil-off rate. The storage vessel is surrounded by a copper radiation shield which, like the shields surrounding the magnet, is cooled by a small fraction of bypassed helium vapor. All components in addition are copiously insulated with evacuated aluminized mylar superinsulation. Typical liquid helium loss figures for the components are $\sim 0.4 \ell h^{-1}$ for the storage vessel and up to $4.5 \ell h^{-1}$ for the umbilical and magnet. The reason for this relatively large loss is the thinness of the insulation around the magnet, particularly in the front of the pole piece. The storage vessel, umbilical and magnet are hydraulically supported on a trolley so that smooth vertical and horizontal adjustments of the system are possible. Figure 4 illustrates the system schematically.

A 300A programmable DC power supply, protection circuits, and dump resistor, liquid helium level interlocks, magnet charging rate controls, and the readout system for a portable Hall probe are mounted in a separate transportable module.

Although we built the magnet originally to induce clotting in cerebral aneurysms injected with a suspension of iron microspheres suspended in albumin, it has since been used successfully in somewhat different applications. Rather than rely on the thrombosing action of the iron, we use vulcanizable silicone doped with carbonyl iron to control its placement. The method is simple: A catheter is inserted via the appropriate artery and positioned within the region to be occluded in the diseased organ. An injection catheter is fed through the larger catheter until the former protrudes a small distance. The magnetic field is switched on. A small amount, usually not more than 3 mℓ, of the ferrosilicone mixture, consisting of carbonyl iron microspheres suspended in Silastic diluted with liquid silicone, is injected with an infusion pump. Just prior to the injection stannous octoate is added to catalyze the vulcanizing action. At body temperatures the vulcanizing action is very rapid so that the magnetic field can be switched off after 20-30 minutes. Should the initial occlusion be inadequate, the process can be repeated several times.

The magnet has now been in use for several years during which time various studies [10] have effectively demonstrated not only that ferrosilicone vascular occlusion can be used on cerebral aneurysms but that it has many potential applications in tumor therapy. As long as a major arterial blood supply to the tumor, as distinct from the host organ, exists, the solid tumor in principle can be destroyed by this technique.

We are currently designing a new, totally self-contained, magnet system, shown diagrammatically in Figure 5. Although the magnet, a niobium-titanium solenoid with a central field of 7.5T, is rigidly attached to the helium supply vessel, the dewar is capable of up-down motion, rotation, and about 15^{o} of swing. The total height does not exceed 2 m when fully extended. A nitrogen jacket is included. All the necessary services, such as the vacuum system, power supply, monitoring and safety instrumentation, and the hydraulic system, are contained in the console. The magnet is designed to be cooled down and serviced in the laboratory: it can be wheeled into the operating room or angiographic center and charged when required.

CONCLUSION

We have briefly reviewed the use of magnetic fields in medicine and surgery, and we have indicated the usefulness of guidance and retentive techniques. We hope that this rather new methodology will stimulate further interest so that the marriage of a relatively recent technology with mankind's oldest science will lead to yet more powerful methods of combating disease.

ACKNOWLEDGEMENTS

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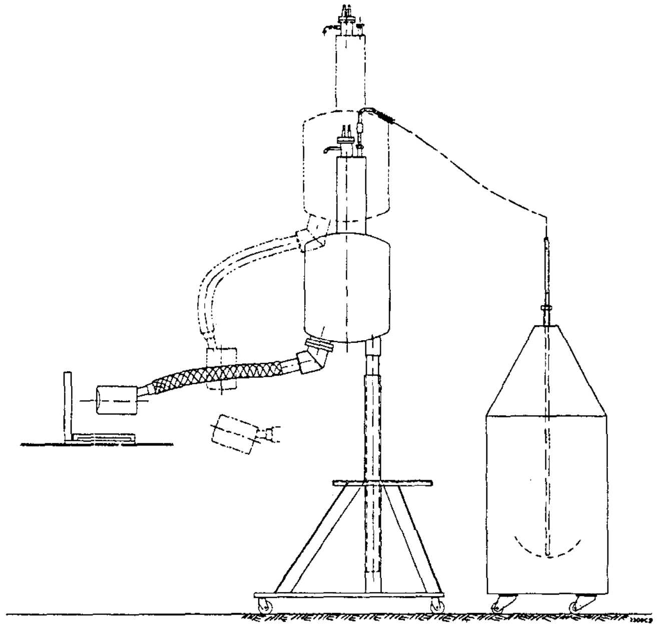

Fig. 4--Arrangement of the SLAC superconducting magnet for use with the ferromagnetic occlusion technique.

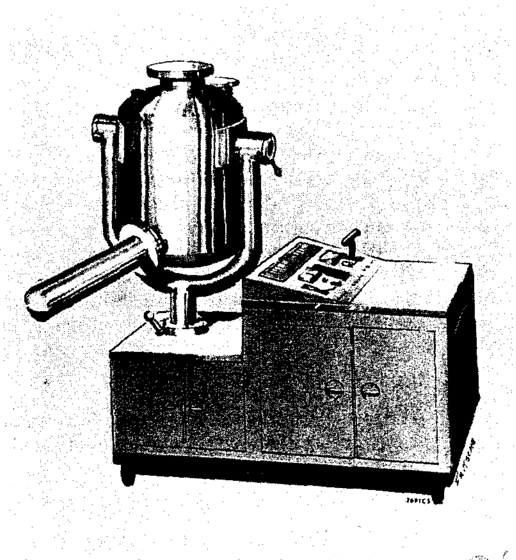

Fig. 5--General arrangement of a new superconducting magnet system for magnetic surgery.

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