Project name:	3D Printed coil
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Research motivation:

In communication with the cardiology department of the Regional Hospital in Liberec, IKEM in Prague and other medical workplaces, it was stated that the pacemakers (PM) used in some cases show malfunction and the reason cannot be determined from the stored data. Doctors' suspicions are directed at external interference – although the pacemakers are sufficiently shielded, as is the wire connecting the electrode, the tip of the electrode, which is used to monitor heart activity, is no longer shielded (see Pictorial attachment). This sensor reads values in the order of tens of millivolts, and it is therefore likely that the external electromagnetic field (EMG) will cause an order of magnitude higher interference; noise of 3 - 10 Hz is particularly dangerous, e.g. far-field EGM (interference under high voltage, etc.)

Technology used:

The aim of the project was to verify the possibilities of fully three-dimensional printing of measuring coils. PCB 3D printing technology (Dragonfly LDM 2.0 by Nanodimension) enables the production of PCBs in a completely new way. The existing Multilayer PCB production technologies allow the individual turns of the coil to be layered on the layers, and thus the Sequential Built Up (SBU) technology from the innermost layers by adding additional layers of prepreg and metallic foils; Production takes place in the following steps:

1. Etching of the innermost layers of the motif.
2. Applying prepreg and Cu foil on both sides and pressing.
3. Drilling and plating buried via, filling holes and reinforcing the motif.
4. Etching of the motif of the other two layers of the coil.
5. Repeat steps 2-3-4 until the total number of layers required is reached.

It is requiring the repeated involvement of a number of production machines from mechanical to chemical lines, and in addition to the production of common waste in the form of chemicals and wastewater, it is also necessary to take into consider a large technological waste in the form of the removal of the technological environment of the motif.

The process developed by us uses a single machine for production and does not generate any waste except for the liquid for cleaning the print nozzles, the production consists of uploading the print data and starting the printing procedure. The same data can be used as print data as in the production of traditional technology (Gerber + Excellon) from a traditional eCAD design system, but because the main advantage of eCAD, i.e. the car router, cannot be used in the case of specialized and highly sophisticated motifs, it is possible to use a traditional 3D CAD modeling tool when preparing print data. In this case, we can also use spatial generators, taking full advantage of the possibilities of the 3D space intended for placing the motif, in this case the spatial coil.

The coil under development has been designed for use in medical applications, for "wearable" Holter electronics, for this reason it is essential that the final product is harmless to health. In cooperation with the manufacturer, biocompatibility according to EN ISO 10993-5, conformity according to DIN EN ISO/IEC 17025, health safety is also stated by the UL certificate 14847193. The products can thus be used in both in-vitro and in-vivo medical applications and are suitable for medical clinical research.

Coil development:

The possibility of interfering with the pacemakers used (PaceMaker = PM) is zero, PM is used in the patient's body for 5 years or more, and in addition, there are relatively few patients with malfunctions, so after consulting with doctors, we decided to develop a sensor that would be used as an external "wearable" device and that would scan the surrounding field; the result would then be compared with the PM data, and the doctor would then be able to see when interference has occurred and could either recommend how to behave (avoid a disturbing environment) or adjust the PM settings.
Disposable ECG probes were selected to determine the dimensions, which medical staff can work with:

Fig. 1 - Demonstration of the appearance of disposable ECG probes

The development of the coil was performed in the CAD system Autodesk Fusion 360 in the form of two models - one for the conductive motif, the other for the insulator, where both models are moved to the identical origin of the coordinate system and "subtracted" from each other as solids before generating the production data (STL format). The resulting STL data was exported to the proprietary Flight tool, which calculated the print file "apcb" / "pcbjc", which are intended for the printing process itself, i.e. the control of printing in individual printing layers (slices) using CI (conductive) and DI (dielectric) inks. During the development, several variants of coil geometry and their arrangement were verified. In parallel with the 3D design, the parameters of individual variants were verified in the Ansys simulation tool.

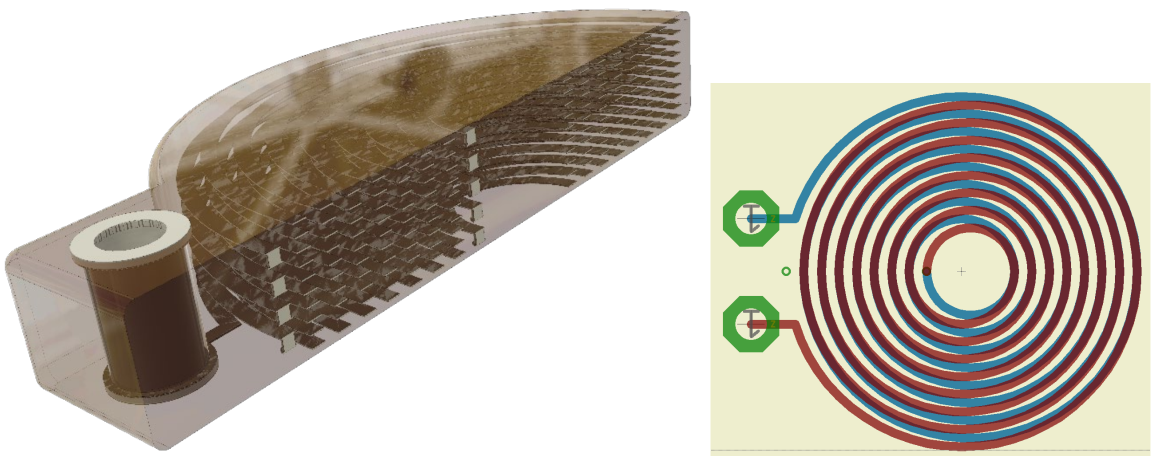


Fig. 2 - Sensor model

Description of the resulting construction:

Number of conductive layers	**10**
Number of threads in a layer / total	**8 / 8x10**
Total sensor thickness	**2 mm**
Sensor outer diameter	**10 mm**
Conductor and insulation gap widths	**250 µm**
Wire thickness	**17 µm**
Insulation gap in the Z-axis	**200 µm**

The dimensions used are not borderline, they were used mainly to verify function, manufacturability, production process and, last but not least, to verify the computational model. It is therefore possible to further increase the number of "layers", the number of turns, etc. in the same space.

Theoretical Background for Design of Sensor:

List of vector definitions:

$\mathbf{B}$ – ext. magnetic induction	[T]
$\mathbf{\Phi}$ - total magnetic flux	[V.s]
$\mathbf{E}$ – external electric field strength	[V/m]
$\mathbf{H}$ - outer magnetic field strength	[A/m]
$\mathbf{S_i}$ - partial area of induction loop	[m^2]
$\mathbf{S_t}$ - total effective area of the induction loops	[m^2]

List of scalar definitions:

$u_{it}(t)$ - voltage induced at the sensor	[V]
$\mu=\mu_r\mu_0$ - absolute permeability	[H/m]
S_{sj} - partial area of induction loop, spiral solution	[m^2]
S_{st} - area of induction loop, 1layer, spiral solution	[m^2]
S_{ct} - area of ind. loop, 1layer concentric circles	[m^2]
S_t - total effective area of the induction loops	[m^2]

The alternating external electromagnetic field is transformed into the induced electric field $\mathbf{E_i}$ in the sensor. The product of electric field strength $\mathbf{E_i}$ indicates induced voltage U_i at the output contacts of the sensor. We suppose that the external magnetic field has a low dynamic in the range from 1Hz to 100Hz. Maxwell´s Equations in differential and integral form (1, 2) and Faraday's Induction Law can be used to deduce the induced electric field. These principles can offer optimal solutions to our problems and be suitable for electromagnetic circuits of sensors that we consider electric circuits with parameters in 3-dimensional space.

$$\mathrm{rot}\,\mathbf{E} = -\frac{d\mathbf{B}}{dt}\left[\frac{V}{m}\right] \quad \mathbf{B} = \mu\mathbf{H}\left[\frac{V}{m}\right] \qquad (1)$$

$$\oint_C \mathbf{E}\partial r = -\frac{d}{dt}\left(\iint_S \mathbf{B}_n \partial\mathbf{S}\right) \quad [V] \qquad (2)$$

$$u_i(\mathrm{t}) = -\frac{d\mathbf{\Phi}}{dt} = -\frac{1}{dt}(\mathbf{B}\partial\mathbf{S})[V] \qquad (3)$$

The quasi-stationary case of the equations (1,2,3) is a simplified fundamental theory. The use of circuit theory simplifies many calculations, especially in our case. We define this problem mathematically as nonstationary boundary value assignment. The nonstationary electromagnetic field is solved in a bounded, geometrically limited space significantly more significant than the sensor's dimensions with the boundary at a defined specific time interval. The design of the sensor is based on the principle of superposition. The total magnetic flux $\mathbf{\Phi_t}$ is given by the sum of the partial magnetic fluxes passing through the individual loops and coils [4].

$$u_{it}(\mathrm{t}) = -\frac{1}{dt}\left(\sum_{\mathbf{1}}^{\boldsymbol{n}} \boldsymbol{\Phi}_{\boldsymbol{i}}(\mathbf{t})\right)[V] \qquad (4)$$

n number of spiral coils

We postulate that $\mathbf{B}$ is constant magnetic field is homogenous in space of sensors.

$$u_{it}(\mathrm{t}) = -\frac{\mathbf{B}(t)}{dt}\mathbf{S_t} = -\frac{\mathbf{B}(t)}{dt}\left(\sum_{\mathbf{1}}^{\boldsymbol{n}} \mathbf{S}_{\boldsymbol{i}}\right)[V] \qquad (5)$$

We need to achieve the maximum number of coils in one layer. The limitation is the minimal width of insulation distance between wires and their width. The principle of Archimedes spiral is more valuable than concentric circles in PCB design. It is used only one crossing between layers. The coils of single layers are serial-connected. In the present case it is necessary to assume that the magnetic field passing through the sensor is not perpendicular to the sensor surface $\mathbf{S_t}$.

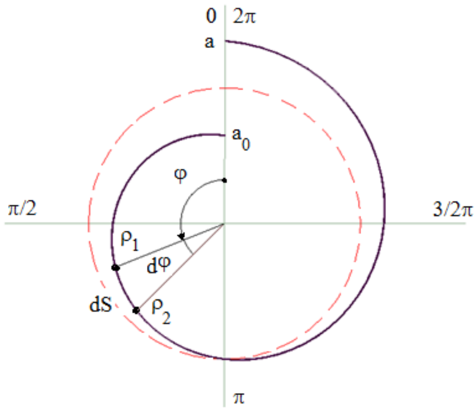


Fig. 3 - The mathematic principle of Archimedes' spiral

a_0	distance of initial value of spiral, $\varphi=0$
a	coefficient of spiral pitch, $a>0$
ρ_1, ρ_2	radius vectors
dS	surface of spiral sector
φ	angle of turning in a polar coordinate system
$d\varphi$	difference angle of turning

$$\rho(\varphi) = a_0 + \frac{1}{2\pi}a\varphi[m] \qquad (6)$$

The counting of the surface sector of the spiral is by approximately the sector of a circle. The S_{st} represents the total effective area of the spiral in one layer.

$$dS \sim \frac{{\rho_{1,2}}^2}{2}\, d\varphi \quad [m^2],\ \rho_{1,2} = \frac{\rho_1+\rho_2}{2} \quad [m] \qquad (7)$$

$\Delta\varphi \to 0$

$$S_{st} \sim \sum_1^k \frac{{\rho_k}^2}{2}\, d\varphi_k \quad [m^2] \qquad S_t = lS_{st}[m^2] \qquad (8)$$

$$\rho_k \leq a_0 + na \qquad [m] \qquad (9)$$

$\Delta\varphi \to 0, k \to \infty$ (Integer)

n	number of coils in one layer
l	number of conductor layers

The similar result we obtained using simplified by calculation of concentric circles. The sum of concentric circles (dashed line) affords sufficient estimation of the total surface of magnetic flow.

$$S_{ct} \sim \sum_0^n \pi \left(a_0 + \frac{1}{2}a + na\right)^2 \; [m^2] \tag{10}$$

$$S_t = lS_{ct}[m^2] \tag{11}$$

The induced voltage $u_{it}(t)$ of the sensor is given by the equation (5), [3]

Final measurements:

Finally, the properties of the sensor were measured in reality and these measurements confirmed the correctness of the Ansys model:

Serial inductance L_S = 27 µH, (calculated at 27.5 µH) – for frequencies from 100Hz to 100kHz);
Series resistance R_S = 23.6 Ω.

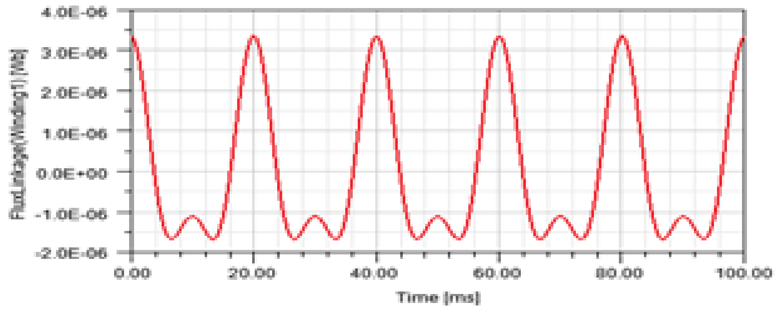


Fig. 4 - Waveform of the input field on the sensor surface

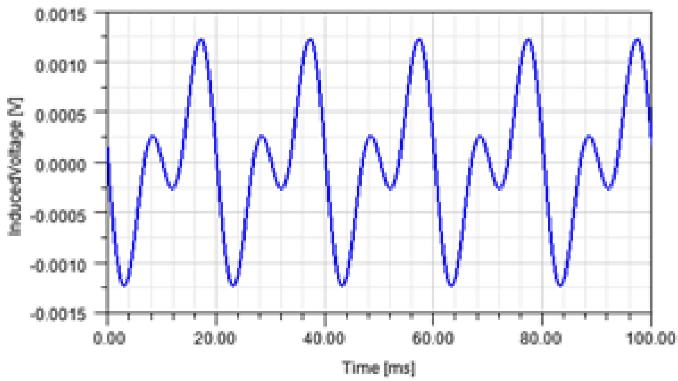


Fig. 5 - Voltage waveform on the sensor

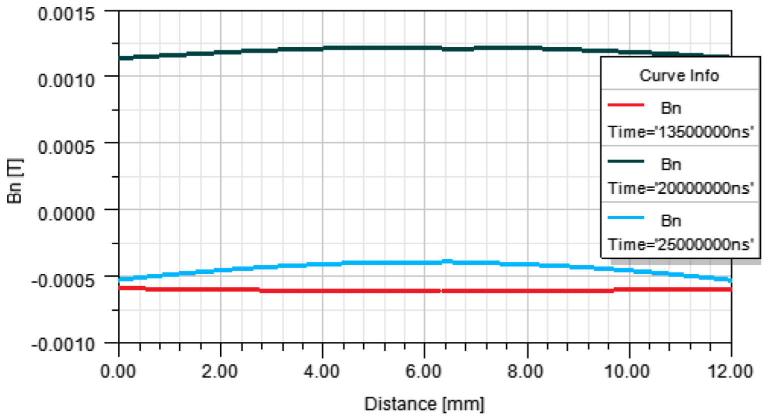


Fig. 6 - MG induction course on the sensor surface

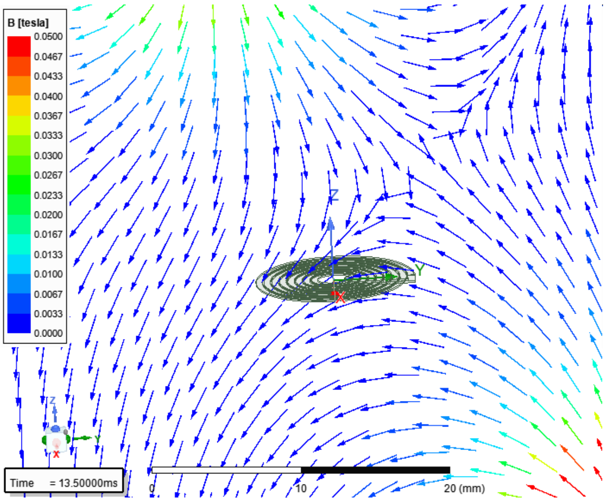


Fig. 7 - Pattern vectors – Ansys FEM model

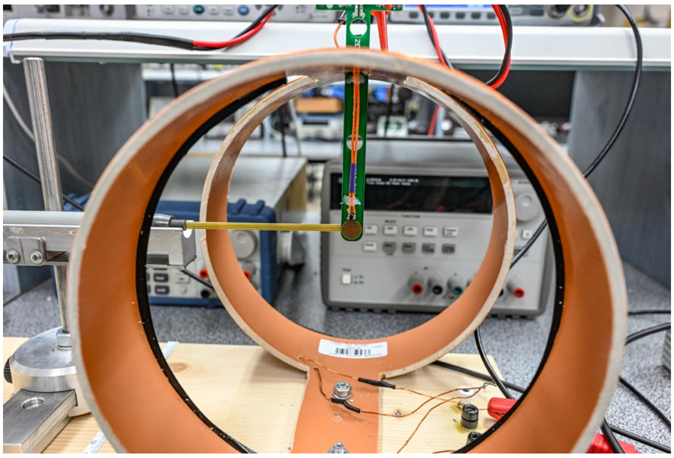


Fig. 8 - Measuring workplace

Evaluation of the result:

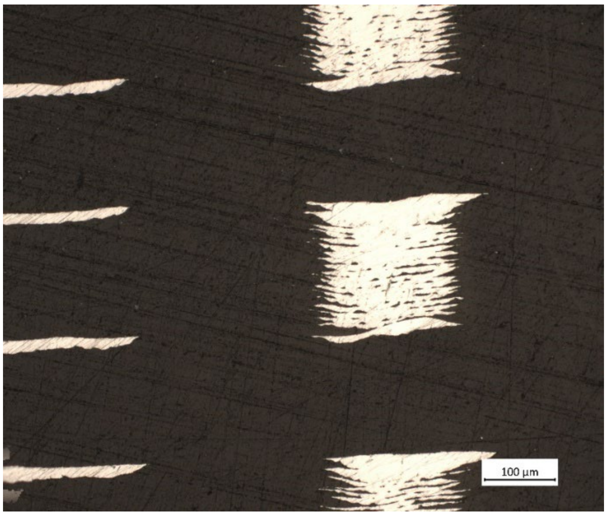
The developed functional sample fully confirmed the suitability of 3D printing of PCBs for these types of tasks. AME technology offers completely new possibilities for the implementation of classic linear circuits, especially in miniature design; The higher conductor density of coils made in this way allows for better inductive and capacitive parameters with smaller dimensions and higher energy density. At the same time, it brings the possibility of using 3D space, abandoning the "standard" layer system, implementing buried plugged via (or design without via). The use of the entire 3D space will allow further improvement of the properties of such constructed sensors, opening up completely new possibilities of use in biophysics, especially in clinical research, as linear sensors of electromagnetic fields in a wide range of frequencies and amplitudes. The resulting sensors are suitable for direct use in clinical research, in the development of new medical devices. The measurements confirmed that the AME outputs correspond very well to the results obtained in discrete FEM modelling systems, e.g. in ANSYS, Comsol, etc. A controller for practical use is currently being developed; once the function is verified outside the laboratory environment, the design of the 3D printed sensor in the form of the second generation will also be optimized for real-world use.

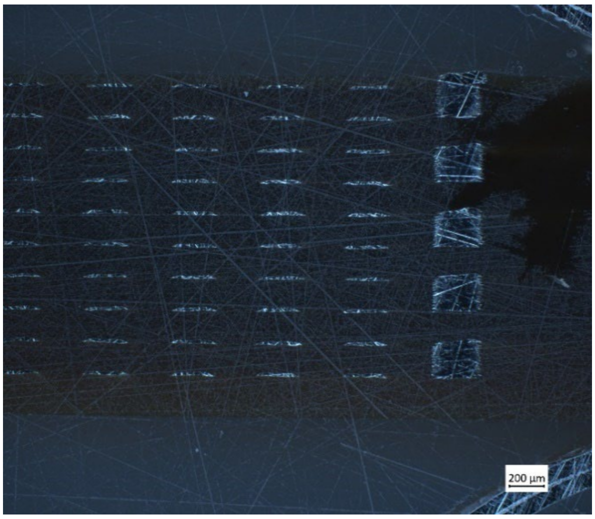
Presentation of results:

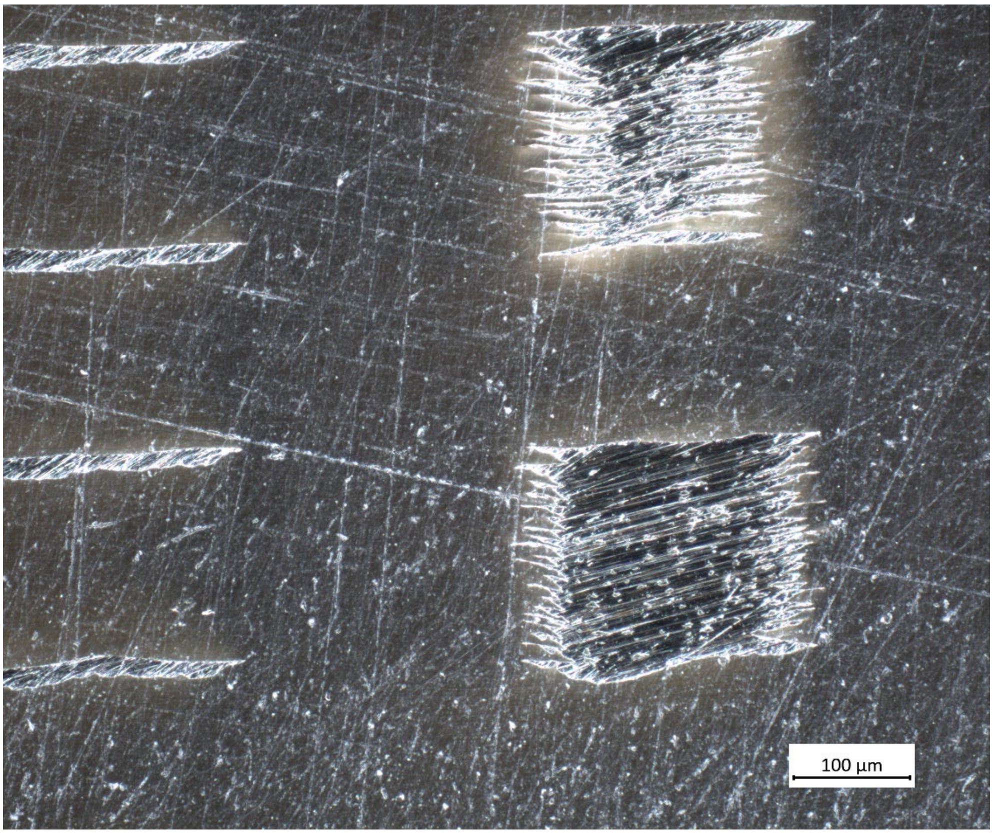
Richter, A., Z. Ferková, L.Petržílka, Z. Plíva: *The Inductive External Low-frequency Sensor of Electromagnetic Field using AME 3D PCB Nanotechnology for Biophysics Applications*. 2024 IEEE 21st International Power Electronics and Motion Control Conference (PEMC). Pilsen, Czech Republic, September 30 - October 3, 2024. ISBN: 979-8-3503-8522-9/24

Pictorial attachments:

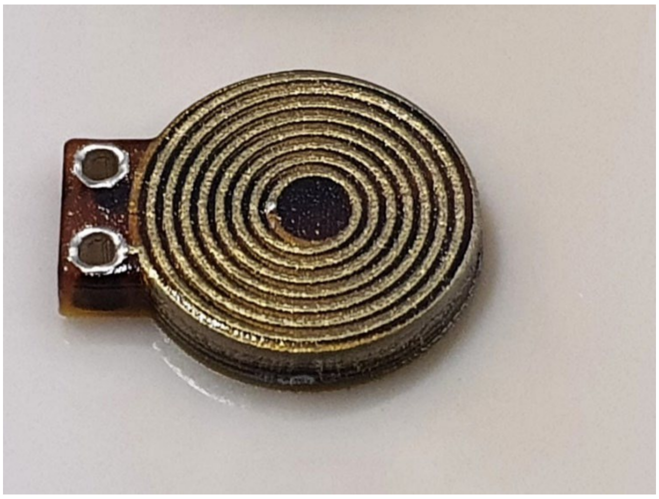
Sensor metallographic grinding:







Sensor photos:

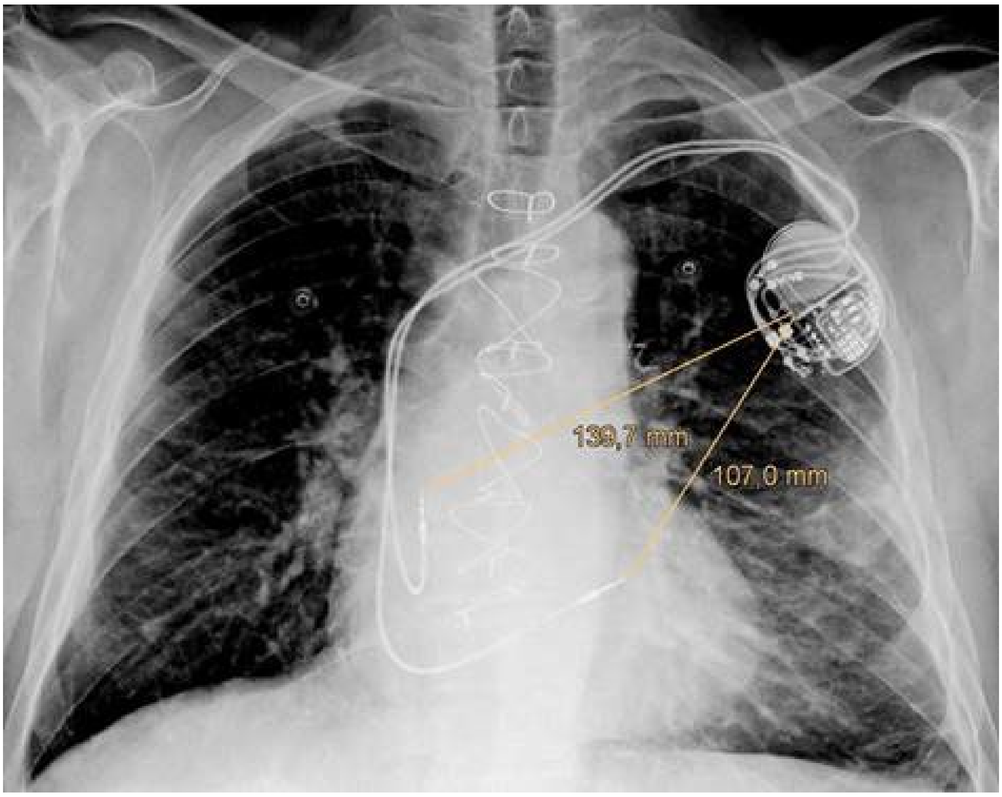


Note: The perimeter of this test specimen has been abraded to make the internal structure visible.

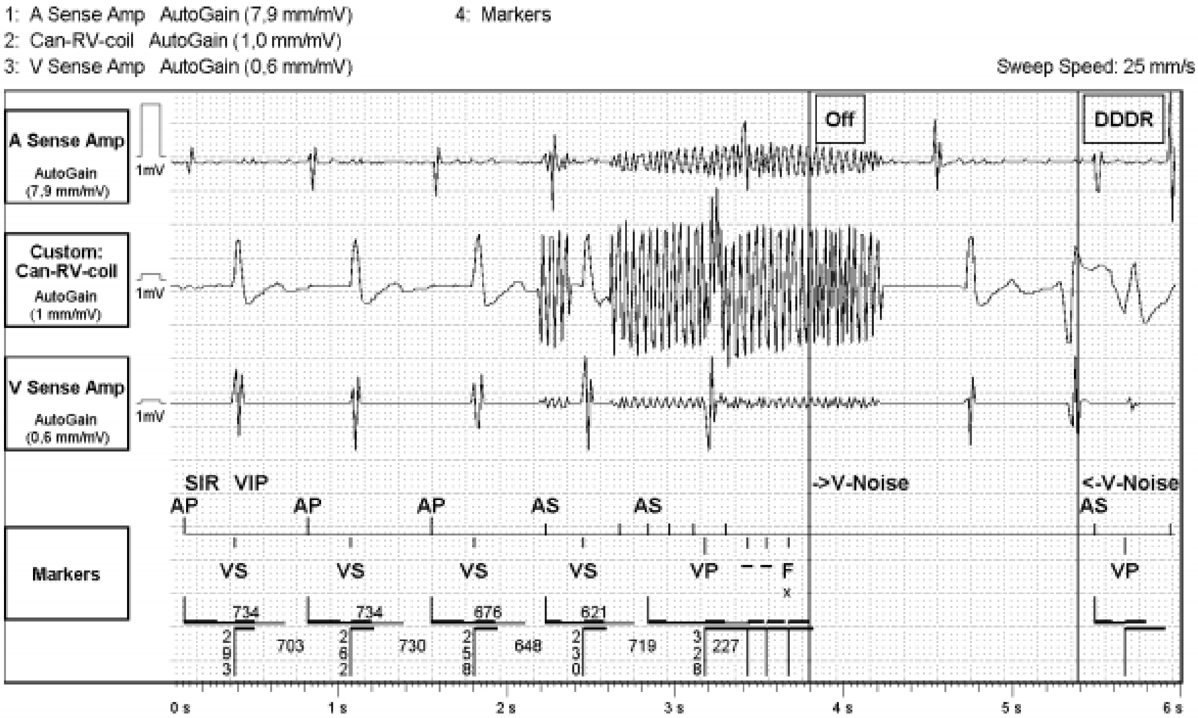
Application of the sensor in clinical research:

The pacemaker itself and the electrode leads are shielded. The problematic input of electromagnetic interference with the pacemaker is the contact of the electrodes with the heart muscle. The action potential of the heart muscle can be in the range of units of mV. Based on the X-ray images taken at the time of pacemaker implantation in the patient, we obtain a spatial idea of the position of the system in the patient's chest and the exact orientation of the induction loops in different configurations. The ECG recording of the pacemaker will only show us the presence of interference, but not its nature in the amplitude and time domain. If this happens repeatedly in a patient, e.g. during occupation, long-term monitoring (e.g. 24 h) is one way to localize the source of EMI. The research is conducted in cooperation with the Cardio Centre KNL, Regional Hospital Liberec.

J. Morava, J. Kupec, A. Richter, T. Souček, *Holter Ecg Monitoring as Method for Assessing Interaction of Implanted Pacemaker And Source of Electromagnetic Interference*, Lekar a technika – Clinician and Technology, Czech Society for Biomedical Engineering and Medical Informatics, p. 18-22, 5 pages, ISSN: 0301-5491, n. 1, [Online], 2022
J. Morava, A. Richter, J. Kupec, T. Souček, L. Slavík, *Interaction of work electromagnetic field with implanted cardiostimulation system, analysis by in vitro method at patient practicing of profession*, Health and Technology, Springer Verlag, p. 801-807, 7 pages, ISSN: 21907188, n. 4, [Online], 2022



Anatomy of pacemaker placement in the patient



Detection of electromagnetic interference at the electrodes of an implantable cardiac device – The standard ECG waveform obtained from the pacemaker is shown in line 2. The signal from the in vitro electrodes is disturbed from about the 2nd second of the recording. Due to the interference, which lasts for 4 heartbeats, the heart activity cannot be evaluated. The pacemaker goes into a slower synchronous mode and returns after the interference has stopped. The character of the interference cannot be determined from the pacemaker data recording.