# CAE Environment for the Efficient Development of Clinical MRI Scanners

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## Abstract

In this paper a CAE environment for a more efficient development of clinical magnetic resonance imaging (MRI) scanners is demonstrated. The modeling scheme is based on a finite element method (FEM) and allows the full 3D calculation of the electromagnetic, mechanical and acoustic fields including their couplings. In order to compute the complex models, efficient algorithms based on iterative solvers utilizing advanced preconditioners and algebraic multigrid methods are used. A recent addition allows the modeling of the electric field sources based on an application of the law of Biot-Savart. Therewith, a finite element model of the complex coil structure can be avoided. This simplification not only reduces CPU time but also modeling time and complexity enormously. Furthermore, the numerical simulation scheme has also recently been combined with automated optimization algorithms. These use a powerful implementation of a deterministic, numerical optimization algorithm, which is based on a robust SQP-method in combination with an efficient methodology to calculate the gradients of the objective and constraint function. With this set-up a computer based optimization of the design of MRI scanners with respect to different parameters, such as eddy current losses and emitted noise, is made available.

## 1 Introduction

To reduce the efforts in the development of clinical magnetic resonance imaging (MRI) scanners, precise and efficient computer modeling tools have to be used. With these computer simulations, the costly and lengthy fabrication of different prototypes, required in optimization studies by conventional experimental design, can be avoided. At present, computer modeling tools, which are based on finite and boundary elements, are well established only in the design of either pure electromagnetic or mechanical-acoustic field problems. However, complex interactions of coupled physical fields still can not be efficiently treated by commercially available finite element codes. Therefore, in this paper a CAE environment for a more efficient development of clinical magnetic resonance imaging (MRI) scanners is demonstrated.

## 2 Governing physical behavior

In the case of clinical MRI scanners, especially, the prediction of the coupled magnetomechanical-acoustic behavior is of increasing interest for the MRI industries, as shown below.

Fig. 1 a.) displays the cross-section of a typical clinical MRI scanner. Two basic com-

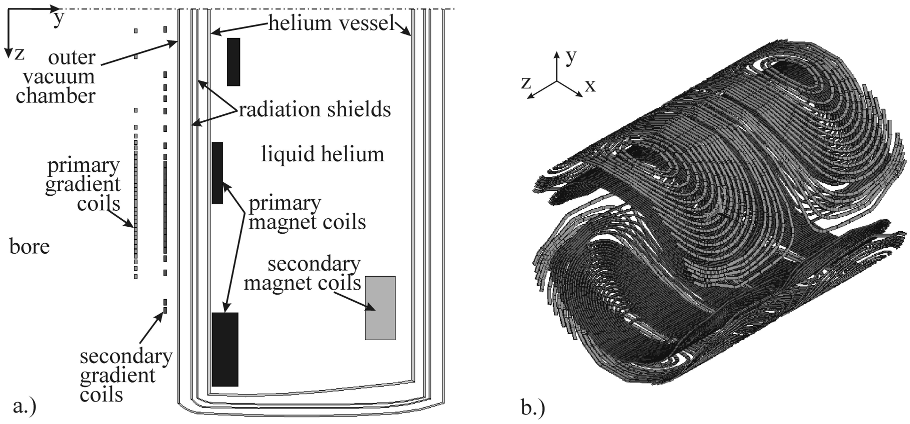


Figure 1: a.) Cross-section of a typical clinical MRI scanner, b.) Schematic of a transverse gradient coil (y-gradient)

ponents of the MRI system, which are of interest in this paper, are the main magnet and the gradient coils. The main magnet generates a strong static magnetic field in the bore to align the nuclei in the patient tissues [1]. The gradient coils produce highly linear magnetic field gradients within the imaging volume for the spatial localization of the magnetic resonance signal and the selection of the slices. There are three sets of windings, designed to generate a gradient along three orthogonal axis (x,y,z). The z- gradient is excited with a pair of rotational symmetric coils with essentially Helmholtz geometry. For the generation of the transverse gradients x and y, four 3D saddle coil structures carrying the same current are assumed, as displayed in Fig. 1 b.). In MRI scanners with superconducting magnets, the gradient coil is mounted close to the cryostat (see Fig. 1 a.)). The cryostat consists essentially of the stainless steel outer vacuum chamber, aluminum radiation shields, and the stainless steel helium vessel. During clinical use, the gradient coil is driven with an appropriate choice of pulse sequences. Therefore, the gradient coil vibrates due to the Lorentz forces, acting on its wires. Additionally, even with a well-shielded gradient coil, a small residual magnetic stray field is generated resulting in induced eddy currents in the surrounding electrically conductive shields of the main magnet. Furthermore, the interaction between the magnetic stray field and the strong static magnetic field of the magnet (up to 3T) results in a very complex response function due to the load by the gradient coil, governed by strongly coupled Lorentz forces, mechanical vibrations, and eddy currents.

Therefore, for the computer simulation of a clinical MRI scanner, the following physical fields including their mutual couplings have to be modeled: magnetic field, mechanical field, acoustic field, coupling 'Magnetic Field - Mechanical Field' and Fluid-Solid interaction. The governing partial differential equations describing this magnetomechanical-acoustic system have already been reported [3] and will not be repeated here.

# 3 Computation scheme

In this paper, the governing equations are solved using the finite element (FE) method (FEM). The modeling scheme allows the full 3D calculation of the electromagnetic, mechanical, and acoustic fields including their mutual strong couplings. In order to reduce the CPU-time of the 3D simulations we have applied efficient solvers based on enhanced pre-conditioners and algebraic multigrid (AMG) methods. In the 3D calculations an AMG solver is used to solve the magnetic field equations, whereas a GMRES algorithm including advanced preconditioning techniques is applied to the mechanical/acoustic system. This combination of solvers has proven to supply a stable solution of the fully coupled transient simulation and yielded a clear strategy of how to handle our kind of physical problem more routinely from now on [3]. For a detailed discussion of the theory of the underlying finite element scheme, we refer to [2].

In the initial implementation as presented in [3] the conductors of the transverse gradient coil were discretized very fine by finite elements. Consequently, a large number of magnetic finite elements as well as a high modeling effort was necessary in the simulation of the MRI-scanner. Furthermore, the optimization of the gradient coil and the superconducting magnet with respect to eddy current losses and the emitted noise were done manually. Therefore, the modeling scheme [3] has been extended by the modeling of the electric field sources based on an application of the law of Biot-Savart (see Section 3.1) and the combination with automated optimization algorithms (see Section 3.2).

## 3.1 Utilizing Biot–Savart's Law

For further reduction of CPU-time and modeling effort, the conductors of the gradient coil are replaced by an equivalent set of line currents. The resulting magnetic field intensity $\mathbf{H}_{\mathrm{GC}}$ can be computed in advance by applying Biot–Savart's law

$$\mathbf{H}_{\mathrm{GC}}(x',y',z') = \frac{1}{4\pi\mu_0} \oint_C \frac{I_{\mathrm{GC}}\, d\mathbf{s}\times\mathbf{r}}{r^3}\,, \tag{1}$$

with $(x',y',z')$ the field (observation) point, $I_{\mathrm{GC}}$ the equivalent line currents of the gradient coils, and $\mathbf{r}$ the distance to the observation point. Therewith, we decompose the magnetic field $\mathbf{H}$ into the magnetic field $\mathbf{H}_{GC}$ of the gradient coils and the magnetic field $\mathbf{H}'$ of the superconducting magnet coils as well as the reaction field given by $\mathbf{H} = \mathbf{H}_{\mathrm{GC}} + \mathbf{H}'$. In addition, we also have to decompose the magnetic vector potential. Therewith, the formulation for the magnetic vector potential $\mathbf{A}'$ reads as follows

$$\boldsymbol{\nabla}\times\left(\frac{1}{\mu}\boldsymbol{\nabla}\times\mathbf{A}'\right) = \mathbf{J}_{\mathrm{e}} - \boldsymbol{\nabla}\times\mathbf{H}_{\mathrm{GC}} - \gamma\dot{\mathbf{A}}'\,. \tag{2}$$

where $\mathbf{J}_\mathrm{e}$ denotes the free current density, $\mu$ the permeability, and $\gamma$ the electrical conductivity. With this decomposition, we can compute in a first step the magnetic field $\mathbf{H}_\mathrm{GC}$ generated by the gradient coils using (1), and in a second step solving (2).

### 3.2 Combining with automated optimization

For a more user-friendly application of the CAE-tool described above, the numerical simulation scheme is combined with automated optimization algorithms. These use a powerful implementation of a deterministic, numerical optimization algorithm, which is based on a robust SQP-method in combination with an efficient methodology to calculate the gradients of the objective and constraint functions [4].

A general nonlinear constraint optimization problem with optimization parameters $z \in \mathbb{R}^n$ is written as

$$\min_{z \in \mathbb{R}^n} \quad \Phi(z), \quad \text{s.t.} \quad g_i(z) \le 0 \ (i = 1, \ldots, m) \tag{3}$$

The objective function $\Phi(z)$ contains the eddy current losses, for example, and is written as follows:

$$\Phi(z) = \int_{f_1}^{f_2} \omega(f) \Big( \int_\Omega \rho \mathbf{A}^2 \mathrm{d}\Omega - Q_{Target} \Big)^p \mathrm{d}f \tag{4}$$

$Q_{Target}$ denotes the desired upper level of eddy current losses in a frequency range $[f_1, f_2]$. To improve the flexibility of the objective formulation, we have introduced a weighting function $\omega(f)$. The nonlinear constraints $g(z)$ are needed to ensure the compliance of practical important coil design criteria, such as the gradient strength and nonlinearity (NL) in the field-of-view, inductivity, shielding, power dissipation, etc. ...

For the solution of problem (3) an efficient mathematical optimization algorithm (SQP) has been employed, which requires the gradients of the objective $\Phi(z)$ and constraint functions $g_i(z)$ $(i = 1, \ldots, m)$ with respect to the design vector $z$ in each iteration of the optimization process. Within the CAE-Tool those gradients are provided by means of semi-analytical methodologies during coupled magnetomechanical-acoustic field computation.

## 4 Verification of the 3D computation scheme

The initial verification of the 3D magnetomechanical computation scheme has already been reported in [3] and will not be repeated here. In this paper, a three-dimensional (3D) magnetomechanical finite element model has been set up for the modeling of the transverse gradient induced eddy current losses (see Fig. 2). In the full FE- model the current-loaded conductors of the gradient coil and the magnet coil are discretized using pure magnetic coil elements. The cryostat as well as a small ambient region are modeled using magnetomechanical elements based on the moving-mesh method [2]. Finally, the outer surrounding air is modeled by pure magnetic elements. In this dynamic analysis an AMG solver has been used to solve the magnetic field equations, whereas a GMRES algorithm including an advanced ILU(4) pre-conditioner (fill-in 4) has been applied to the mechanical system [3].

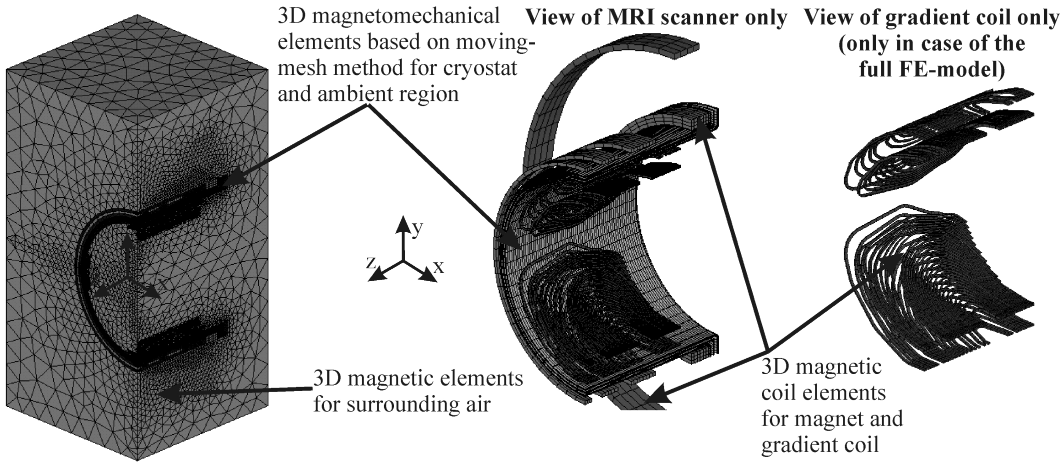


Figure 2: 3D magnetomechanical finite element model of a MRI scanner with transverse gradient coil (y-gradient)

In order to function properly, the conductors of the transverse gradient coil and, therefore, the surrounding air has to be discretized very fine. Consequently, a large number of magnetic finite elements as well as a high modeling effort is necessary in the simulation of the MRI-scanner (see Fig. 2). Therefore, a coarser FE-model has been applied recently, in which the 3D magnetic coil elements for the gradient coil were eliminated completely. The electromagnetic excitation of the gradient coil is now realized by an equivalent set of line currents (see Section 3.1). Therewith, the complexity of the computer model as well as the required computer ressources are reduced tremendously as shown in Tab. 1.

Table 1: Comparison of computer ressources of the full FE-model and the coarse FE-model utilizing Biot-Savart's Law, Note: Opteron 248-2200MHz computer

| model | finite elements | unkowns | physical memory | CPU/time step |
|---|---|---|---|---|
| full FE-model | 290000 | 450000 | 4900 MByte | 5.6 minutes |
| coarse FE-model | 164000 | 290000 | 4100 MByte | 3.6 minutes |

The coarse FE-model utilizing Biot-savart's Law described above has been verified by comparing simulated eddy current losses inside the helium vessel with the corresponding results of the full FE-model as well as with measured data (see Fig. 3 a.)). Considering that the deviations of subsequent measurements of the eddy current losses were within a range of $\pm 15$ %, an acceptable agreement between both simulation results and measured data was achieved. In a second step, our optimization algorithm has been verified by comparing the automatically optimized eddy current losses with the corresponding manual data (see Fig. 3 b.)). Here, the CAE-tool was extended to compute the derivatives of the objective function during the magnetomechanical field computations by means of semi-analytical methodologies. In comparison to a black box approach based on finite differences, we obtained a problem dependent speed up factor between 5 and 10. Already after five iterations a significant improvement of the design could be achieved. In Fig. 3 b.) the objective $\Phi(z)$ of the tenth optimization step is plotted. The result, found by automated optimization, was very close to the solution, obtained by manual improvements. Furthermore the optimization history gave a clear indication, which coil design criteria, formulated as constraints in $g_i(z)$, were contradictory.

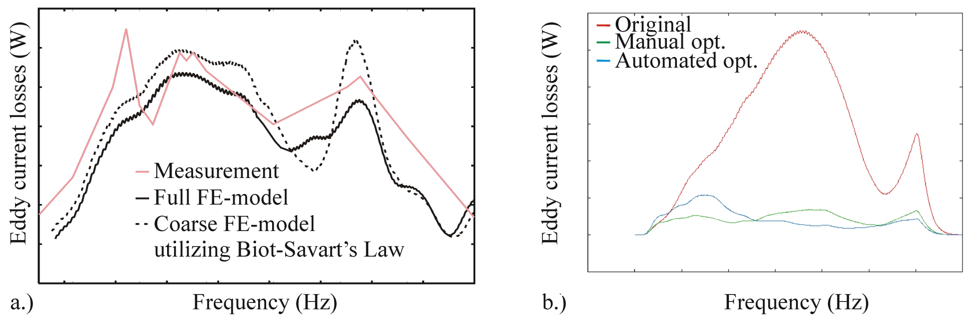


Figure 3: Verification of magnetomechanical field computations: a.) Comparison of measured and calculated eddy current losses, b.) Comparison of manually and automatically optimized eddy current losses

## 5 Application in an industrial computer-aided design process

To explore the practical applicability of the developed simulation scheme in an industrial computer-aided design process, the superconducting magnet and the gradient coil were numerically analyzed and optimized with respect to the gradient induced eddy current losses in its cryostat and the emitted noise. For the modeling of the transverse gradient induced eddy current losses, the three-dimensional (3D) magnetomechanical finite element model described in Section 4 has been setup. In the course of this automated computer-optimization, the knowledge of this numerical study was put into our new prototype to reduce the eddy current losses in the cryostat. As can be seen in Fig. 4 a.), significant smaller eddy current losses were obtained with the optimized system. Furthermore, the numerically predicted improvements of the eddy current losses could be successfully confirmed by measurements on the new prototype. Furthermore, to demonstrate the usability of our sound field computations in an industrial computer-aided design process, the sound emission of the MRI scanner was numerically analyzed and optimized. For a detailed discussion of the corresponding computer model we refer to [3]. As can be seen in Fig. 4 b.), the numerically predicted improvements of the sound emission could be successfully confirmed by measurements on the new prototype.

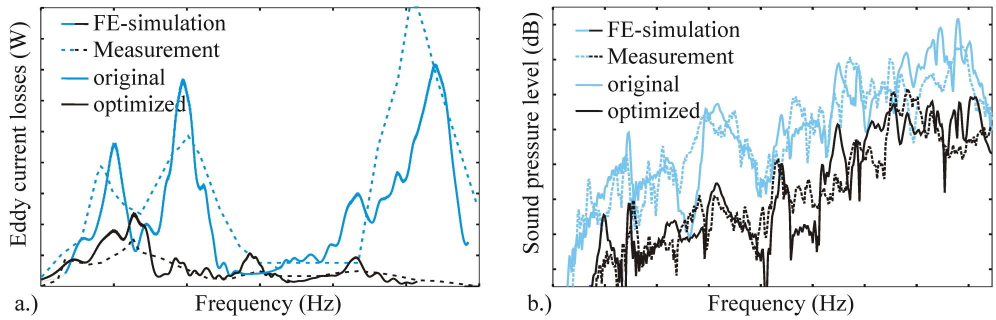


Figure 4: Computer-optimization of the MRI-scanner a.) with respect to eddy current losses, b.) with respect to emitted noise

# 6 Conclusions

A numerical scheme based on a 2D/3D finite element method has been developed for computer modeling of the coupled magnetomechanical-acoustic behavior of a real clinical magnetic resonance imaging (MRI) scanners. In order to reduce the CPU-time of the 3D simulations we have applied efficient solvers based on enhanced pre-conditioners and algebraic multigrid (AMG) methods. For further reduction of CPU-time and modeling effort, the finite elements for the conductors of the gradient coil are replaced by an equivalent set of line currents. Furthermore, the numerical simulation scheme is combined with automated opimization algorithms. The developed simulation scheme is well suited to the industrial computer-aided design of clinical MRI scanners, since an optimization with a significant reduced number of prototypes can be achieved.

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