Gradient Performance and Gradient Amplifier Power

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When talking about gradient performance, most will only state the maximum amplitude (strength) of the gradient system and the maximum slew rate (speed). But actually, there is much more to a gradient system. There are numerous design criteria for a gradient coil to be considered. In addition, there is the gradient power amplifier (GPA) that 'drives' the gradient coil. The performance of the gradient system and the power of the gradient amplifier are closely related.

It is the intention of this paper to discuss the key characteristics of the gradient coil as well as the gradient amplifier, with their inter-dependencies and their clinical benefits.

We will start with the gradient amplifier as it is the 'driving force' of the gradient system.

Characteristics of a gradient power amplifier

The gradient power amplifier (GPA) is characterized by:

A. Peak current (I_{max} in A)
B. Peak voltage (U_{max} in V)
C. "Gradient amplifier power"
D. Long-term performance of the amplifier
E. Other 'intricacies'
F. Special two-amplifier design

There are three axes of the gradient amplifier, one each for the gradient coils in x, y, and z directions.

A. Peak current (I_{max} in A)
The peak current I_{max} of the gradient amplifier (measured in amperes, A) has a direct relation to the max. gradient amplitude of the gradient system, to be discussed in detail in the next chapter.

Typical peak currents of today's commercially available whole-body gradient systems are in the range of 100 A to 1,200 A.

B. Peak voltage (U_{max} in V)
The peak voltage U_{max} of the gradient amplifier (measured in volts, V) has a direct relation to the maximum slew rate of the gradient system, also to be discussed in the next chapter.

Typical peak voltages are in the range of 500 V to 2,250 V.

C. Gradient amplifier power
A very good measure for the integral gradient amplifier power (P_{amp}) is the product of current and amplitude:

$$P_{amp} = I_{max} \cdot U_{max} \qquad \text{(Eq. 1)}$$

The apparent amplifier power P_{amp} (measured in kVA or MVA) directly relates to the integral gradient performance, discussed in the next chapter. For the sake of simplicity, we will call the 'apparent power' just 'power' hereafter. For the differentiation between apparent and real power, see [1].

The (apparent) power of today's gradient amplifiers is extremely high. The strongest currently available gradient amplifier has a power of 1,200 A x 2,250 V = 2.7 MVA. The three amplifiers for the three gradient axes add up to 8.1 MVA. For visualization purposes: 8.1 megawatts (MW)[1] correspond to more than 11,000 horsepower, i.e. roughly a dozen Formula-1 racing cars.

D. Long-term performance of the amplifier
A high-power gradient amplifier generates a lot of heat. An efficient cooling is required to guarantee stability and prevent overheating. It should be noted that the typical time constants of the gradient amplifier to heat up (<< 1 sec) are much shorter than those of the gradient coil (that takes much longer time to heat up, in the order of minutes).

In order to achieve highest-possible long-term performance, today's state-of-the-art gradient amplifiers are water cooled.

E. Other intricacies
There are many more important characteristics for the quality of a gradient amplifier, such as accuracy, linearity, and stability. But these go beyond the scope of this paper.

F. Special two-amplifier design
There are some gradient systems in the market from one vendor that feature two sets of gradient amplifiers. There

[1] The apparent power of the gradient amplifier (measured in kVA) is different from the real power (measured in kW). The equalization of apparent and real power (kVA vs. kW) and subsequent translation into horsepower is not correct. It is only stated to indicate the very high power of the gradient amplifiers in everyday language.

are two gradient amplifiers for each gradient axis that can be switched in parallel or in serial mode:

i. The parallel mode results in high current (= high amplitude), but low voltage (= low slew rate).

ii. The serial mode results in high voltage (= high slew rate), but low current (= low amplitude).

This possibility of switching offers a certain flexibility to adjust the gradient performance to the specific needs of the application (high amplitude or high slew rate). The disadvantage is that the whole imaging sequence can only be run in either of the two modes; a switching during the sequence is not possible. This is for example relevant in diffusion-weighted imaging with EPI (echo-planar imaging) since the diffusion encoding requires high amplitude while the EPI readout requires high slew rate.

The consequence of this switching is that the imaging sequence can only use (i) lower slew rate or (ii) lower amplitude. Neither the gradient coil efficiency nor the integral gradient performance (see paragraph 3 in the next chapter) are improved; it is only possible to choose between two different setups.

Characteristics of a gradient coil

1. A gradient coil is characterized by numerous aspects:
2. Sensitivity and peak amplitude (G_{max} in mT/m)
3. Inductance and peak slew rate (SR_{max} in T/m/s)
4. Efficiency and "integral gradient performance"
5. Winding density
6. Linearity
7. Inner diameter
8. Coil thickness
9. Shielding
10. Force compensation
11. Heat generation in the coil
12. Cooling efficiency of the coil
13. Other "intricacies"
14. Special two-coil design

There are three axes of the gradient coil, in x, y, and z directions. Each axis is powered by a separate gradient amplifier (see previous chapter).

The points 1 and 2 define the 'integral (peak) gradient performance' (3), while the winding density (4) can be used to trade off amplitude versus slew rate. The points 5–9 are characteristics of the gradient coil with their own benefits or drawbacks that also have an influence on the peak performance (1–3). Finally, the points 10–11 define the long-term performance of the gradient coil. All these points will now be discussed in detail.

1. Sensitivity and peak amplitude
The peak amplitude (G) is the strength of the gradient system, i.e. the steepness of the magnetic field. It is measured in mT/m.

The gradient amplitude (in combination with the RF excitation pulse) defines the slice thickness. The integrals of the gradient pulses in phase-encoding and readout directions define the in-plane resolution of the image. In general, the amplitude is important for the spatial resolution of the image. The amplitude is especially critical for diffusion imaging, as the diffusion weighting (b-value) depends on the square of the gradient amplitude.

The highest peak amplitude in a commercially available whole-body 3-Tesla MR system is 80 mT/m. This means that the magnetic field at full gradient strength varies over the maximum field of view of 50 cm (± 25cm) by ± 0.02T. This looks like a small change only in comparison to the static field B_0 of 3T, but see the gradient amplifier chapter how much power is required for this gradient field.

When looking at *k*-space coverage, the gradient amplitude defines the speed with which *k*-space is traversed, e.g. going from left to right in *k*-space during the readout of an echo.

For a given gradient coil, the gradient strength is directly proportional to the current (see amplifier chapter above):

$$G = \eta I \qquad \text{(Eq. 2)}$$

G is the gradient amplitude; I is the current in the coil, generated by the gradient amplifier; η is the coil sensitivity[2] that describes how much gradient strength one gets per ampere of current.

The coil sensitivity is proportional to the winding density (number of coil turns) of the gradient coil (see Annex for more details). It also depends on many of the characteristics of the gradient coil, such as inner diameter and linearity. The sensitivity is typically in the range of 0.05 mT/m/A (mT/m per ampere).

2. Inductance and peak slew rate
The slew rate (SR) is often referred to as the speed of the gradient system. It is measured in T/m/s (or mT/m/ms). It describes how fast a certain gradient amplitude can be switched on, starting from 0 mT/m. It is related to the rise time (T_{rise}), i.e. the time it takes to achieve certain gradient amplitude, by:

$$SR = G / T_{rise} \qquad \text{(Eq. 3)}$$

[2] The 'sensitivity' is also called 'efficiency' in the literature. However, we will use the term 'efficiency' differently, as an integral measure of the gradient coil, taking into account the relation of both amplitude and slew rate in comparison to the amplifier power.

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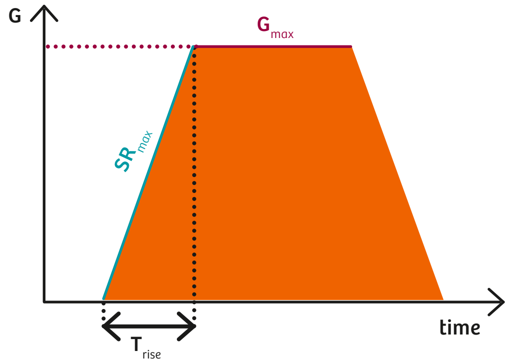

Figure 1: Gradient amplitude (G), slew rate (SR), and rise time (Trise).

In earlier times, the rise time was specified instead of the slew rate. However, the slew rate is a metric that can be better compared between different gradient systems. In this context, it is only important to make sure that the max. amplitude and the max. slew rate can be used simultaneously, i.e. can be combined in a single pulse. For this, the additional specification of the rise time (from 0 to max. amplitude) is valuable for cross-checking.

The gradient slew rate is especially important for fast sequences, such as fast gradient echo (GRE), TrueFISP, and echo-planar imaging (EPI). Especially single-shot EPI depends on the max. slew rate almost exclusively since it is typically used with low spatial resolution (matrix size 64–256) where the max. gradient amplitude is not reached. The speed of the readout pulses and the phase-encoding 'gradient blips' depends only on the slew rate. Note, however, that diffusion EPI requires high gradient amplitude for the diffusion encoding.

The highest peak slew rate in a commercially available whole-body MR system is 200 T/m/s. This means that the rise time from 0 to maximum gradient strength of, for example, 45 mT/m is 225 µs (see Eq. 3).

When looking at *k*-space coverage, the gradient slew rate defines the *acceleration* in *k*-space, i.e. how fast a certain *speed* in *k*-space (related to the gradient amplitude, see paragraph 2) is achieved.

For a given gradient coil, the gradient slew rate is directly proportional to the voltage (see amplifier chapter above):

$$SR = \eta / L \cdot U \qquad \text{(Eq. 4)}$$

SR is the gradient slew rate; U is the voltage generated by the gradient amplifier; η is the sensitivity of the gradient coil (see paragraph 1); L is the inductance of the gradient coil.

The resistance of the gradient coil also plays a role in the calculation of the slew rate; however, this effect is only approx. 5% and we therefore disregard it here. This is discussed in more detail in the Annex.

The 'transmission' from voltage to slew rate is inversely proportional to the winding density (number of coil turns) of the gradient coil (see Annex for more details).

3. Efficiency and integral gradient performance

The two key characteristics of the gradient system, amplitude and slew rate are often written in the simplified (dimensionless) form "G_{max}/SR_{max}". For example, "45/200" means a max. amplitude of 45 mT/m and a max. slew rate of 200 T/m/s.

The product ($G_{max} \cdot SR_{max}$) is a good measure for the (short-term) "integral gradient performance" of the gradient system:

$$\text{Gradient Performance} = G_{max} \cdot SR_{max} \qquad \text{(Eq. 5)}$$

For the sake of simplicity, we will only write the dimensionless number, without the (rather clumsy) unit "mT/m · T/m/s = mT²/m²/ms". As an example, the MAGNETOM Prisma with 80/200 gradients has a gradient performance of "16,000" (per axis).

For a given gradient coil, the gradient performance is proportional to the gradient amplifier power ($I_{max} \cdot U_{max}$):

Gradient Performance ∞ Amplifier Power

Using Equations 2 and 4, this yields:

$$G_{max} \cdot SR_{max} = \eta^2/L \cdot I_{max} \cdot U_{max} \qquad \text{(Eq. 5)}$$

The "conversion factor" between amplifier power and gradient performance we will call the "**efficiency**" of the gradient coil, ε:

$$\varepsilon = \eta^2/L \qquad \text{(Eq. 6)}$$

So we can write Eq. 5 as:

$$G_{max} \cdot SR_{max} = \varepsilon \cdot I_{max} \cdot U_{max} \qquad \text{(Eq. 7)}$$

We see that the amplifier power directly relates to the gradient performance of the MR system.

There are a number of other characteristics of the gradient coil, however, that influence the efficiency ($\varepsilon = \eta^2/L$) between amplifier power and gradient performance. These will be discussed now.

4. Winding density

The density of the coil winding (current density) can be used to trade off gradient amplitude versus slew rate.

A higher density will increase gradient sensitivity, i.e. "more amplitude per current". At the same time, higher density will increase inductance, i.e. "less slew rate per voltage".

This is proportional. As an example, if the density is doubled, gradient amplitude will be two times higher, but slew rate will be halved. If the density is halved, gradient amplitude will be halved, but slew rate will be two times higher. The integral gradient performance ($G_{max} \cdot SR_{max}$, Eq. 5) will remain identical, independent of the winding density.

For the sake of simplicity, we are disregarding the low ohmic losses in the gradient coil. For a more detailed discussion, see the Annex.

5. Linearity

The linearity (also sometimes called homogeneity) of the gradient coil describes the deviation from an ideal linear 'steepness' of the magnetic field. The linearity (actually, the non-linearity) is measured in %.

A low linearity of the gradient coil will result in image distortions and in a smaller maximum field of view (FOV). The distortions can be corrected by software means (with a distortion correction algorithm applied on the reconstructed images), but this comes at the expense of lower spatial resolution in the periphery of the field of view, due to an interpolation involved in the 'expansion' of collapsed voxels. Also the b-values in diffusion imaging will become spatially dependent due to non-ideal linearity.

However, a lower linearity will increase the efficiency (as defined in paragraph (3)) of the gradient coil with respect to max. amplitude and max. slew rate. This means that a higher "integral gradient performance" ($G_{max} \cdot SR_{max}$, as defined above) can be achieved with the same gradient amplifier power if linearity is sacrificed.

Finally, a lower linearity will result in a lower level of peripheral-nerve stimulations (PNS), i.e. more gradient performance can be used without stimulating the patient.

Fig. 2 shows the typical shape of the gradient linearity of two different gradient coils [2]. The x-axis shows the distance from isocenter (± z). The y-axis shows the resulting magnetic field from the gradient amplitude, $B = G_z \cdot z$. The steepness of the lines is the gradient strength. A lower linearity will result in a smaller maximum FOV. The strongest 'elevation' of the field (per time, i.e. dB/dt), indicated by the rectangles, is a measure for the PNS limit. The dotted line indicates a gradient system with higher gradient strength (larger steepness) but lower linearity. Since the maximum elevation is not higher (same B_{max}), the PNS limit only occurs at higher gradient strength than with the gradient system with lower gradient strength. Alternatively, it would be possible to use a higher slew rate (shorter rise time) at the same amplitude. Typically,

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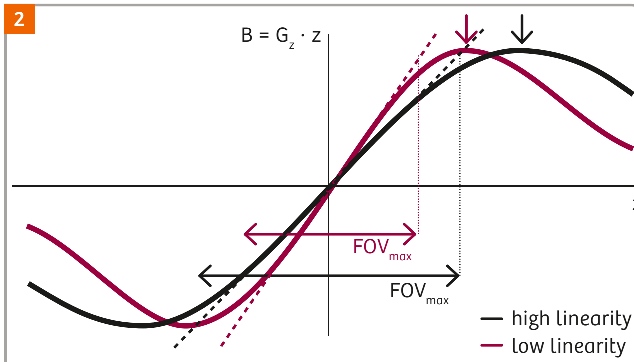

Figure 2: Typical shapes of the gradient linearity for two different gradient systems (black: high linearity; red: low linearity). Solid lines: actual shape of the magnetic field; dotted lines: ideal linear gradient field; FOV_{max} denotes the maximum possible field of view with acceptable deviations from linearity. Arrows indicate the locations with maximum elevation of the magnetic field (outside the linearity volume).

the location of maximum elevation is approx. 10 cm outside of the max. FOV, e.g. at ± 35 cm for a max. FOV of 50 cm.

In the end, the linearity of the gradient coil needs to be carefully balanced to achieve the best compromise between conflicting measures:

- High linearity decreases geometric distortions and therefore increases the maximum FOV.
- Lower linearity increases the efficiency of the gradient coil and also lowers the level of PNS.
- The gradient linearity should also be balanced with the magnetic-field homogeneity B_0 which also has an influence on the max. FOV. There is little use of combining high linearity with low magnet homogeneity and vice versa.

Here is an extreme example of how much performance (efficiency) can be gained by sacrificing linearity:

The MAGNETOM Symphony with Quantum gradients (introduced in 1999) had 30/125 gradients, i.e. a gradient performance of 3,750 (as defined in Eq. 5). It was powered by a gradient amplifier with 380 A and 2,000 V, i.e. gradient power = 0.760 MVA.

The MAGNETOM Sonata (introduced in 2000) featured the same magnet as well as many other commonalities. It had 40/200 gradients, i.e. a gradient performance of 8,000. It was powered by a gradient amplifier with 500 A and 2,000 V, i.e. gradient power = 1.000 MVA.

If we set these values into relation, we see that the MAGNETOM Sonata had 2.13 times the gradient performance of the MAGNETOM Symphony with Quantum gradients, although it had only 32% higher amplifier power. The efficiency gain, so to speak, was 2.13/1.32 = 1.62, i.e. 62% more gradient performance per amplifier power.

This was achieved by sacrificing gradient linearity (and also with some compromise regarding shielding, see below). Although both systems featured the same magnet, i.e. had identical magnet homogeneity, the MAGNETOM Symphony offered a full 50 cm FOV, while the MAGNETOM Sonata was restricted to a max. FOV of only 40 cm – based on the lower gradient linearity.

An even more extreme example are dedicated head insert gradient coils. For the max. FOV in head imaging in the range of 20 cm, much lower linearity is required, and much higher gradient performance can be achieved with the same gradient amplifier power.

These are extreme examples. The differences in linearity of today's whole-body MR systems will be much smaller, resulting in efficiency differences (gradient performance per amplifier power) in the range of max. 10–15%.

6. Inner diameter

The inner diameter of the gradient coil equals the patient bore diameter plus 2 times the thickness of the RF body coil. See Figure 3 for visualization.

The inner diameter of the gradient coil is the most critical factor for the integral gradient performance (i.e. the efficiency as defined in paragraph 3), as this is inversely proportional to the 5th power (!) of the diameter [3]:

$$\varepsilon \propto 1 / \text{Diameter}^5 \qquad \text{(Eq. 8)}$$

If we assume a typical thickness of the RF body coil of 3 cm, a system with 70 cm patient bore (i.e. 76 cm inner diameter of the gradient coil) in comparison to a system with 60 cm patient bore (i.e. 66 cm inner diameter of the gradient coil) will have $(66/76)^5 \approx ½$ the gradient performance (with the same amplifier power). In other words, a 70 cm system requires twice (!) the amplifier power to achieve the same gradient performance as a 60 cm system.

As an example: MAGNETOM Prisma (G = 80 mT/m, SR = 200 T/m/s) has the same gradient amplifier as MAGNETOM Vida[3] with XQ gradients (G = 45 mT/m, SR = 200 T/m/s).

The difference of 80 mT/m vs. 45 mT/m is similar to the factor of 2, consistent with theory.

And here we are talking only about the peak performance. The long-term performance is another aspect, to be discussed below.

7. Coil thickness

A state-of-the-art gradient coil is actively shielded, i.e. it consists of a primary coil for the generation of the gradient field (actually 3 coils for x, y, and z) and a secondary coil for the shielding.

Figure 4 shows the different layers of a gradient coil [4]. Going from the inside out: The primary coils (x, y, z) generate the gradient field. They are water cooled (the exact design and the number of cooling layers may vary among different gradient coils; this is a simplified visualization only). Some MR systems feature an additional coil set for patient-specific high-order shimming [5]. The whole gradient coil body is filled with epoxy to achieve optimal stiffness and stability of the gradient coil. Iron plates are inserted into the passive shim pockets during the installation shimming procedure to improve B_0 homogeneity. The outer shielding coils (x, y, z) reduce the gradient stray field. They are also water cooled.

A thicker gradient coil comes along with a larger distance between primary and secondary (shielding) windings. The shielding will have a (minor) influence on the primary gradient field in the FOV, meaning it will reduce the gradient performance. A larger distance between primary and shielding coil, i.e. a thicker gradient coil, will diminish this effect, thereby increasing gradient performance, i.e. gradient coil efficiency as defined in paragraph 3. Overall, the coil thickness has an influence on the gradient performance, but much less than the r^5 dependency of the inner coil diameter.

The disadvantage of a larger coil thickness is that it requires a larger outer diameter of the gradient coil (with a given patient bore and thickness of the RF body coil). This in turn requires a larger inner diameter of the magnet ("magnet warm bore") which has implications on the magnet design, such as increasing the cost of the magnet. Alternatively, the thickness of the RF body coil can be reduced to provide

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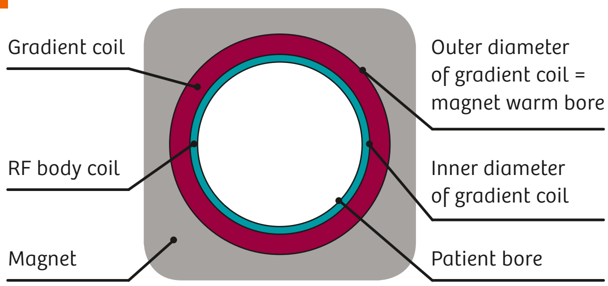

Figure 3: Magnet, gradient coil, and RF body coil

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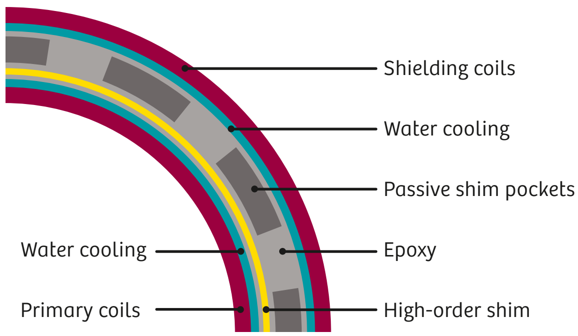

Figure 4: The layers of a gradient coil (not to scale).

[3] 510(k) pending

space for the increase of the gradient coil thickness, but this has negative effects on the efficiency of the RF body coil (more RF power required) and on the specific absorption rate (higher SAR with thinner body coil).

8. Shielding
The shielding coil, positioned outermost in the gradient coil, has the task to minimize the gradient field outside of the coil, thereby minimizing eddy currents in the conductive structures outside the gradient coil, i.e. in the magnet structures (magnet vessel, shields). The shielding factor is measured in %.

On the one hand, a lower shielding factor will result in higher eddy currents which will lead to more artifacts. On the other hand, a lower shielding results in higher efficiency of the coil, i.e. higher gradient performance.

In today's MR scanners, the goal is typically to achieve a shielding factor as close as possible to 100%.

9. Force compensation
Most gradient coils of today's MR scanners are force compensated. This reduces vibrations of the gradient coil, resulting in higher patient comfort as well as higher image quality in critical applications such as diffusion imaging.

However, force compensation comes at the expense of gradient performance (efficiency).

10. Heat generation in the coil
The currents in the gradient coil create heat due to ohmic losses. The generated heat is proportional to the square of the current:

$$\text{Heat Generation} \propto I^2 \qquad \text{(Eq. 9)}$$

This quadratic dependency has severe impacts on the long-term performance of the gradient coil, in particular when considering systems with large patient bore diameters. There is some limit to the maximum voltage of the gradient system, on the one hand because of costs of high-voltage gradient amplifiers, on the other hand to guarantee stability and avoid flashovers.

This means that, for a high-end system that already have high-voltage amplifiers (> 2,000 V), the only means to achieve higher gradient performance is an increase of the current. Remember the r^5 dependency of the gradient performance from Equation 5; a 70 cm system requires approximately twice the amplifier power of a 60 cm system to achieve the same gradient performance if no compromises are made regarding, for example, gradient linearity. An increase of the current by a factor of 2 (with the same max. voltage) will result in 4 times higher heat generation, according to Equation 9. This means that, based on the same gradient technology, a 70 cm system will not have the same long-term performance as a 60 cm system, even if the specified peak performance (G_{max}, SR_{max}) is identical.

A possible solution to the heat problem of large-bore MR systems would be an entirely different gradient technology. In the so-called Connectome gradient systems, different quadrants of the gradient coil are run by separate amplifiers [4], increasing the efficiency of each sub-gradient system compared to a conventional gradient system and thereby reducing heat load. However, such systems are presently not commercially available.

11. Cooling efficiency of the coil
The high heat load of modern high-end gradient systems requires efficient cooling technology. Today's state-of-the-art gradient coils are therefore water cooled. The design of the cooling structures is an important aspect for the long-term performance of the gradient system, for example the number and placement of cooling layers. For most efficient cooling, the cooling layers should be closely attached to the conducting coil structures. There are two different technologies to achieve this:

Some gradient coils in the market make use of hollow tubes, i.e. the cooling water flows in the center of the conductors. This is a very efficient means for cooling; however, it has also a couple of disadvantages:

- The thicker (hollow) wires have a larger minimum curvature radius. This means that the 'fingerprint' pattern of the Golay coils [6] (see Fig. 5), as used for the coils in x and y directions, cannot be as well optimized as with thinner wire structures (that allow tighter curvatures). This can result in less efficient shielding, see paragraph 8. For this reason, hollow wires are mostly used for the z direction only; the solenoid design of the z gradients does not require tight curvatures. But the solenoid z gradient coil is more efficient than the Golay design of the x and y gradients anyway, so the integrated cooling offers least benefit for the z direction.
- The thicker (hollow) wires offer more efficient cooling for static high currents, i.e. static high gradient amplitude. However, they have a much larger cross section. High current changes, i.e. high slew rates, will generate larger eddy currents than in thin wires with small cross section. High slew rates will therefore generate more heat.

The other approach is to manufacture the gradient coil with thin wires. While the cooling with separate cooling structures will not be as efficient for high static currents as

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Figure 5: Golay 'fingerprint' design as used for the x and y gradient coils.

with hollow tubes, the thin wires offer the advantages of lower eddy currents, i.e. less heat generation with high slew rates, and the possibility to optimize the Golay coils with tight curvatures, for optimal shielding.

Both designs have their merits. Our approach is to choose gradient coil wires “as thin as possible” for optimal shielding and lower heat generation with high slew rates, and to optimize the coupling between coil wires and cooling layers for efficient cooling. It should be noted that also the ultra-high-end Connectome gradient coils with gradient amplitudes up to 300 mT/m and slew rates of 200 T/m/s [4] are built with this thin-wire technology.

12. Other intricacies

Although a lot of aspects have been covered above, there is of course more to the design of a gradient coil. There are other intricacies, for example the exact design of the coil windings, the length of the coil, the wire thickness, the connection between primary and secondary coils, etc. All these play a role in the overall gradient performance, but they only have a minor influence compared to the main aspects discussed before.

13. Special two-coil design

There is (or used to be) a gradient system in the market by one vendor that features two sets of gradient coils, driven by the same gradient amplifier:

i. A ‘whole-body’ mode with high gradient linearity (for a large field of view), but with lower gradient performance (lower G_{max}, SR_{max}).

ii. A ‘zoom’ mode with higher performance, but lower gradient linearity (i.e. a smaller FOV).

This possibility of switching offers a certain flexibility to adjust the gradient performance to the specific needs of the application (high performance or large FOV). The disadvantage is that the whole imaging sequence can only be run in either of the two modes; a switching during the sequence is not possible. So, each imaging sequence in a way has to suffer from (i) low gradient performance or (ii) a small FOV with increased spatial distortions. As the two sets of gradient coils also require additional space and make the whole design more complicated, this design with two gradient coils was only developed once (around the year 2000) and not repeated.

Summary

We have seen that there are numerous design criteria for the gradient system – the gradient coil as well as the gradient amplifier. All these criteria influence each other, and a good balancing is required to achieve optimal clinical imaging performance.

We have also seen the importance of the gradient amplifier, being the “driving force” of the gradient system. A high-power amplifier is required to achieve high overall gradient performance – amplitude and slew rate, but also the other characteristics discussed in the paragraphs (5–12), like patient bore, linearity, and shielding. There are no secret technologies to increase the efficiency of a gradient coil. For high gradient performance, high gradient amplifier power is required – or other compromises have to be made.

Table 1 shows a summary of the main characteristics of a gradient coil that have an impact on the peak performance (amplitude, slew rate), with the advantages and disadvantages related to an increase of the respective characteristics.

Gradient Coil Characteristics (increase of)	Advantages	Disadvantages
1. Sensitivity (η^2)	Higher amplitute per amplifier current	Lower slew per amplifier voltage
2. Inductance (L)	Higher slew rate per amplifier voltage	Lower amplitute per amplifier current
3. Efficiency ($\varepsilon = \eta^2/L$)	Higher gradient performance per amplifier power	(Compromises some of the other gradient coil characteristics, see below)
4. Winding density	Trading amplitude vs. slew rate	
5. Linearity	Larger field of view	Lower efficiency
6. Inner diameter	Higher patient comfort	Much lower efficiency (∞ $1/r^5$!)
7. Coil thickness	Higher efficiency	Lower patient bor, or lower efficiency of RF body coil, or larger magnet warm bore required
8. Shielding	Lower eddy currents	Lower efficiency
9. Force compensation	Less vibration (Higher image quality, higher patient comfort)	Lower efficiency

Table 1: Main characteristics of a gradient coil and their respective advantages and disadvantages.

A stronger gradient amplifier will increase the "total performance" of all these characteristics or may diminish the compromises (5–9) that would need to be made for achieving high gradient performance (amplitude and slew rate).

Car Analogy

Here is a car analogy that shows how the performance criteria of a gradient system can be compared to those of a car. The clinical/driving performance describes the benefit, the gradient amplifier / engine is a means to an end, but necessary to achieve a certain performance.

MR Gradient System	Car
Clinical performance: • Gradient strength (G = speed in *k*-space) • Gradient slew rate (SR = acceleration in *k*-space)	**Driving performance:** • Speed • Acceleration
'Engine', means to an end: • Gradient amplifier current (I) • Gradient amplifier voltage (U)	**'Engine', means to an end:** • Power • Torque
'Transmission': • Coil windings: Trading G vs. SR	**'Transmission':** Transmission (gears)

In order to achieve a high performance, a high-power 'engine' is required. For example, a 100-hp car will not be able to reach a maximum speed of 250 km/h (155 mph). In the same way, a weak gradient amplifier will not achieve high performance ($G_{max} \cdot SR_{max}$, Eq. 5), unless severe compromises are made regarding e.g. inner diameter or gradient linearity.

As described above, there are more characteristics to the gradient system (e.g. linearity), as there are more characteristics to a car (e.g. weight). These will influence the relation between "engine" and 'performance'. For example, a lighter car will accelerate faster with the same engine, but at the expense of less safety, less space for passengers and baggage, or similar.

Note on 'Transmission': Most MR systems in the market have a design that corresponds to 'one fixed gear' in a car. Exceptions are the two-amplifier design (see paragraph F) and the two-coil design (see paragraph 13). Both designs might be called 'systems with two-gear transmission'. The disadvantage is that the whole imaging sequence can only be run in either of the gears; 'switching gears' during the sequence is not possible.

Annex

As we have seen in paragraph (1), the gradient strength (for a given gradient coil) is directly proportional to the current:

$$G = \eta I \qquad \text{(Eq. 10 / Eq. 2)}$$

G is the gradient amplitude; I is the current in the coil, generated by the gradient amplifier; η is the coil sensitivity that describes how much gradient strength one gets per Ampere of current.

The coil sensitivity is proportional to the winding density (number of coil turns) of the gradient coil.

We will now derive the relationship between slew rate (SR) and amplifier voltage (U). The voltage in an RL circuit (such as a gradient coil) is given by [7]:

$$U = L \cdot dI/dt + R \cdot I \qquad \text{(Eq. 11)}$$

L is the inductance of the gradient coil. R is the ohmic loss in the gradient coil. dI/dt is the rate of current change (which is related to the slew rate of the gradient system).

The inductance of a typical gradient coil is in the range of 500 µH (micro-Henry). The resistance of a typical gradient coil is in the range of 100 mΩ (milli-Ohm). Inserting the amplifier values of, for example, the MAGNETOM Aera with XQ gradients (I = 900 A, U = 2,250 V, T_{rise} = 225 µs), we see that the term (R · I) is very small in comparison to the first term (L dI/dt), only approx. 5%.

So we can simplify Eq. 11, with only a marginal mistake of approx. 5%, to:

$$U = L \cdot dI/dt \qquad \text{(Eq. 12)}$$

With G = η · I (Eq. 2) and dG/dt just being the slew rate SR, this yields

$$U = L/\eta \cdot dG/dt = L/\eta \cdot SR \qquad \text{(Eq. 13)}$$

$$SR = \eta/L \cdot U \qquad \text{(Eq. 14)}$$

For a solenoid coil, the inductance L is proportional to the square of the number of windings (turns) [8].

Since the coil sensitivity η is proportional to the number of windings N and the inductance L is proportional to N^2, the slew rate SR is inversely proportional to N.

With the amplitude G being proportional to the number of windings N, and the slew rate SR being inversely proportional to N, the gradient performance ($G_{max} \cdot SR_{max}$) is independent of N. So, the number of windings N of the gradient coil (the winding density) can be used to trade off amplitude vs. slew rate. In other words, for a given gradient amplifier with given max. current I_{max} and max. voltage U_{max}, a gradient coil can be designed to balance max. amplitude G_{max} versus max. slew rate SR_{max}. The total gradient performance ($G_{max} \cdot SR_{max}$) will remain the same.

From this, it is also clear that the efficiency of the gradient coil ($\varepsilon = \eta^2/L$, as defined in Eq. 7) is independent of the winding density.

In a publication from 1988, Turner defined a "figure of merit" β of a gradient coil [9,10] as:

$$\beta = \frac{\eta^2 / L}{\sqrt{\frac{1}{V} \int dr \left(\frac{B(r)}{B_0(r)} - 1\right)^2}} \qquad \text{(Eq. 15)}$$

We recognize our efficiency ($\varepsilon = \eta^2/L$) in the numerator. The denominator describes the fractional root-mean-square departure from the required field variation in the region of interest [3]. The denominator is a measure for the non-linearity of the gradient coil. So, we could describe this "figure of merit" as an expansion of our definition of gradient coil efficiency with the gradient coil linearity. This is an even more significant quality measure for a gradient coil than our efficiency. However, no MR vendor specifies this measure for gradient linearity. Our (simpler) definition of gradient coil efficiency ($\varepsilon = \eta^2/L$) relies on published specifications only, and is therefore more practical.

References

1 https://en.wikipedia.org/wiki/AC_power
2 Schmitt F: The Gradient System. Understanding Gradients from an EM Perspective: (Gradient Linearity, Eddy Currents, Maxwell Terms & Peripheral Nerve Stimulation). Proc. Intl. Soc. Mag. Reson. Med. 21 (2013)
3 Hidalgo-Tobon, SS: Theory of Gradient Coil Design Methods for Magnetic Resonance Imaging. Concepts in Magnetic Resonance Part A, Vol. 36A(4) 223–242 (2010)
4 Kimmlingen R: Pioneers of Connectome Gradients. MAGNETOM Flash (68) 2/2017, 122-136
5 Blasche M, Fischer D: Magnet Homogeneity and Shimming. MReadings: MR in RT (3) 2017
6 Golay MJE: Field homogenizing coils for nuclear spin resonance instrumentation. Rev Sci Instrum 1958; 29:313-316.
7 https://en.wikipedia.org/wiki/RL_circuit
8 https://en.wikipedia.org/wiki/Inductance
9 Turner, R: Minimum inductance coils. J. Phys. E: Sci.Instrum. 21:948-952; 1988
10 Turner, R: Gradient Coil Design: A Review of Methods. Magnetic Resonance Imaging, Vol. 11, pp. 903-920, 1993

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