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## Optimal control pulses simulations applied to Magnetic Resonance Elastography for clinical examination with liver iron overload

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### Abstract text

#### Introduction

Magnetic resonance elastography (MRE) is a non-invasive imaging technique enabling quantitative assessment of the mechanical properties of tissues as demonstrated for liver fibrosis [1]. However, technical error rates remain important preventing the development of MRE in clinical use [2].

One of the causes of failure is liver iron overload which induces a decrease in T2 enhancing the need to reduce Echo Times (TE), so as to preserve the NMR signal from the liver.

However, conventional MRE sequences include motion encoding gradient (MEG) which encodes the propagation of shear waves generated by an external actuator in the phase images. Placed between the RF excitation pulse and signal readout, MEG leads to increased TE.

Recent work based on RF pulses designed by using optimal control (OC) theory have shown on preclinical MRI [3,4] that OC-pulses can simultaneously perform slice excitation and motions encoding when applied concomitantly with a constant gradient  $G$ , therefore enabling tremendous TE reduction. However, application to a clinical context has yet to be done. In clinical MRE, the mechanical excitation frequency, maximum amplitudes of gradient and RF pulses ( $BI_{max}$ ) are lower but the amplitudes of mechanical excitations are higher.

In this work, we investigate the possibility to extend the previously proposed OC framework to design motion encoding excitation pulses that are applicable in a clinical context. In particular, the impact of  $BI_{max}$  and the pulse length ( $T_f$ ) is analysed.

#### Methods

The GRAPE (Gradient Ascent Pulse Engineering) algorithm based on gradient descent numerically solves optimal control problems applied to NMR pulse design [5]. It allows computing RF pulses and magnetization trajectories by fulfilling optimality conditions. Based on an initial estimate, the control field is iteratively updated according to the constraints imposed in order to minimise a defined cost function [6] (**Fig. 1, Eq. 1**).

With  $M^{(i,j)}$  the macroscopic magnetization,  $T^{(i,j)}$  the expected target state,  $\rho$  the magnetization amplitude (between 0 and 1) and  $\theta^i = (2\pi x^i)/\lambda$  the magnetization phase which depends on the isochromats' displacement ( $x^i$ ) along the wavelength. Two isochromats separated by a quarter of a wavelength ( $\theta^1 = 0$ ,  $\theta^2 = \pi/2$ ) are considered to obtain a pulse that encodes the wave propagation in the phase images [6].

The slice selection is achieved by applying an RF excitation pulse and  $G$  simultaneously, in presence of the shear wave with frequency  $f_e$  and motion amplitude  $A$ . The resonance offset undergone by the isochromats in the position ( $x^i, z^j$ ), along the propagation direction of the wave ( $z^j$ ) and the slice selection direction (**Fig. 1, Eq. 2**).

Several OC pulses were optimized with clinical MRE parameters. Then, these pulses were propagated through Bloch Equations by simulation. The final obtained transverse magnetization were plotted for each

pulse.

$F_e$  was set to 60Hz and  $A = 1$ mm. The constant gradient  $G$  amplitude was set to 40mT/m in accordance with typical clinical MRI gradient characteristics. Relaxation times were set to  $T1 = 500$ ms and  $T2 = 20$ ms (**Fig. 2**).

The pulse duration was changed from 5 to 25ms by step of 5ms and  $BI_{max}$  from  $10\mu$ T to  $40\mu$ T by step of  $10\mu$ T (**Fig. 3**). Then, for a  $T_f = 25$ ms, a gradual decrease of  $BI_{max}$  from 80 to  $10\mu$ T was carried out ( $10\mu$ T steps) (**Fig. 4**).

## Results

The variation of the pulse length  $T_f$  allowed to see the real impact of this parameter, on the desired target states. Combinations of  $T_f$  and  $BI_{max}$  show that satisfactory optimizations (framed in green) can be reached for  $BI_{max}$  of  $20\mu$ T and above and that for lower  $BI_{max}$ , higher  $T_f$  are required (**Fig. 3**).

For  $10\mu$ T  $BI_{max}$  value (**Fig. 4**), the final states of the isochromat populations are quite far off from the desired target states, the different groups (identified by the three colors) being dispersed on the transverse plane.

## Conclusion

The various simulations showed the influence of  $BI_{max}$  and  $T_f$  on the final states and that  $BI_{max}$  above  $20\mu$ T allows satisfactory convergence. Optimizations with  $T_f > 25$  ms were unsuccessful (not shown here) due to convergence issues. Such results give the framework for the successful application of OC-MRE in a clinical context.

Future work will consist in decreasing the amplitude of the constant gradient, to move away from the maximum limits of the clinical systems and to evaluate its influence on the isochromat final trajectories.

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## Figure captions

**Fig. 1.** Cost Function (Eq. 1) and Resonance Offset (Eq. 2) equations.

**Fig. 2.** Table : Optimization parameters of the different simulations.

**Fig. 3.** Final transverse magnetization states in the transverse plane. Visualization of the different variations of  $T_f$  pulse durations (between 5 to 25ms), for  $B1_{max}$  10, 20, 30 and 40 $\mu$ T. The green boxes highlight satisfactory final states while the red boxes indicate the unachieved target states.

**Interpretation:** Final states of the red and blue isochromats should be close to the plane edges and along the  $\pi/2$  and 0 axes respectively. The black isochromat cluster is expected to be in the center.

**Fig. 4.** Final transverse magnetization states for different  $B1_{max}$  and for  $T_f$  set to 25ms. The ellipses highlight the different isochromat groups.

**Fig. 1**

**Eq. 1:** Cost function equation

$$C(B_1) = \sum_{j=1}^J \sum_{i=1}^N \|\vec{M}^{(i,j)}(T_f) - \vec{T}^{(i,j)}\|^2$$

With  $\vec{M}^{(i,j)}$  the final magnetization,  $\vec{T}^{(i,j)}$  the expected target states:

$$\vec{T}^{(i,j)} = \begin{cases} \rho[\cos(\theta^i), \sin(\theta^i), 0] & \text{if } j \in \Delta z_{in} \text{ (slice thickness)} \\ [0, 0, 1] & \text{if } j \notin \Delta z_{in} \end{cases}$$

**Eq. 2:** Resonance offset equation

$$\Delta B_0^{(i,j)}(t) = G \left[ A \sin \left( -2\pi f_e t + \frac{2\pi x^i}{\lambda} \right) + z^j \right]$$

**Fig. 2**

Figure Reference	$f_e$ (Hz)	A (mm)	G (T/m)	BandWidth (Hz)	$B1_{max}$ Amplitude	$T_f$ (ms)	T1 (ms)	T2 (ms)	
<b>Fig.1</b>	1st line	60	1	0,04	1800	70 $\mu$ T	5,10, 15, 20, 25, 30	500	20
	2nd line	60	1	0,04	1800	80 $\mu$ T	5,10, 15, 20, 25, 30	500	20
<b>Fig.2</b>	60	1	0,02	1800	10 $\mu$ T	25	500	20	
	60	1	0,04	1800	20 $\mu$ T	25	500	20	
	60	1	0,02	1800	30 $\mu$ T	25	500	20	
	60	1	0,04	1800	40 $\mu$ T	25	500	20	
	60	1	0,02	1800	50 $\mu$ T	25	500	20	
	60	1	0,04	1800	60 $\mu$ T	25	500	20	
	60	1	0,02	1800	70 $\mu$ T	25	500	20	
	60	1	0,04	1800	80 $\mu$ T	25	500	20	

**Fig. 3**

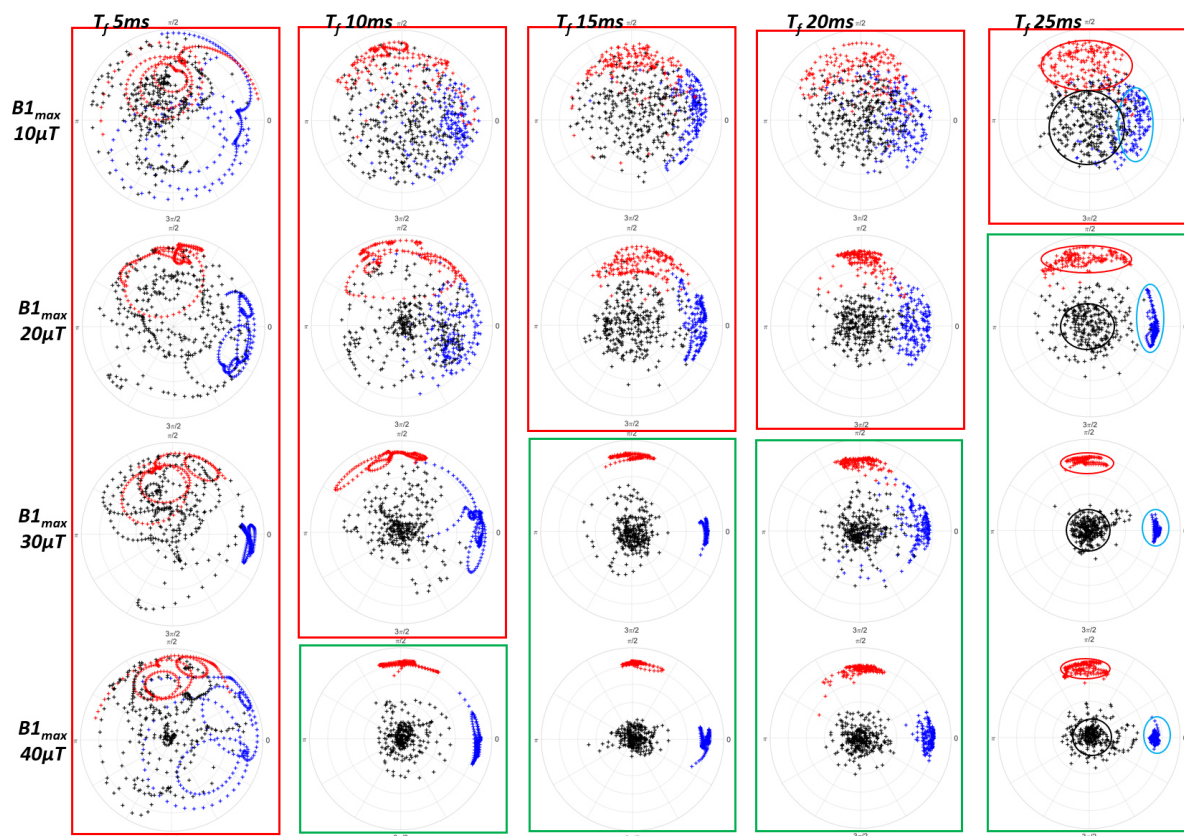


Fig. 4

