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**Dynamic study of blood brain barrier closure after its disruption using ultrasound: a quantitative analysis.**

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3 **Dynamic study of blood brain barrier closure after its disruption using**  
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5 **ultrasound: a quantitative analysis.**  
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## Abstract

Delivery of therapeutic or diagnostic agents to the brain is majorly hindered by the blood-brain barrier (BBB). Recently, many studies have demonstrated local and transient disruption of the BBB using low power ultrasound sonication combined with intravascular microbubbles. However, BBB opening and closure mechanisms are properly understood, especially the maximum gap that may be safely generated between endothelial cells and the duration of opening of the BBB. Here we studied BBB opening and closure under MR guidance in a rat model. First, MR contrast agents (CA) of different hydrodynamic diameters (1 – 65 nm) were employed to estimate the largest molecular size permissible across the cerebral tissues. Secondly, to estimate the duration of the BBB opening, the CA were injected at various times post-BBB disruption (12 min – 24 h). A  $T_1$  mapping strategy was developed to assess CA concentration at the US focal point. Based on our experimental data and BBB closure modeling, a calibration curve was obtained to compute the half closure time as a function of CA hydrodynamic diameter. These findings and the model provide an invaluable basis for optimal design and delivery of nanoparticles to the brain.

**Key words:** *BBB disruption, MR contrast agent, MRI,  $T_1$  mapping, ultrasound, nanoparticles*

## Introduction

Cerebral tissues are isolated from circulating blood by the blood brain barrier (BBB) (Rubin and Staddon 1999). This physiological barrier consists of a lining of tightly packed vascular endothelial cells, different from the peripheral blood vessels (Hawkins and Davis 2005). The tight junctions between these cells restrict the diffusion of microscopic objects (e.g. bacteria) and of large, hydrophilic molecules (>400 Da) from blood to brain parenchyma, while allowing the passage of small, hydrophobic molecules such as O<sub>2</sub>, CO<sub>2</sub>, proteins, and metabolites (Pardridge 2005). Because of the neuroprotective nature of the BBB, delivery of potentially important diagnostic and therapeutic agents is a major challenge in the treatment of most brain disorders. Strategies to design specific drugs targeted to the brain involve finding an efficient mode of drug delivery across the BBB.

Recently, it has been demonstrated that the use of low power focused ultrasound combined with a systemic injection of lipid- (or polymer-) shelled microbubbles enables a non-invasive, local and transient disruption of the BBB (Hynynen *et al* 2001). Many studies were then carried out to (i) establish optimal ultrasound parameters that permit adequate tissular penetration without causing tissue damage (Choi *et al* 2006; Hynynen *et al* 2005; O'Reilly *et al* 2010; O'Reilly *et al* 2011a; O'Reilly *et al* 2011b; Sheikov *et al* 2008), (ii) quantify permeability of the disrupted brain tissue (Vlachos *et al* 2010; Vlachos *et al* 2011), and (iii) evaluate responses to treatments of particular brain disorders including tumors (Chen *et al* 2010; Liu *et al* 2010; Treat *et al* 2009) and Alzheimer's disease (Jordao *et al* 2010; Raymond *et al* 2008). Most of these studies used magnetic resonance contrast agents (MR-CA) for monitoring the processes.

Despite a rapidly growing number of studies, the mechanism of ultrasound-induced BBB opening is understood only poorly. In particular, the maximum space that can be safely generated (ensuring reversibility) between endothelial cells, and the duration for which this

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3 opening lasts, have not been measured. The knowledge of these parameters is crucial for the  
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5 current development of brain targeted nanoparticles. Functionalized MR-CA for instance, are  
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7 available in a wide range of hydrodynamic diameters: gadolinium chelates, a few nm  
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9 (Wadghiri *et al* 2003), iron nanoparticles, 25-100 nm (Mendonca Dias and Lauterbur 1986;  
10  
11 Renshaw *et al* 1986), gadolinium based emulsions or liposomes, 200-300 nm (Devoisselle *et*  
12  
13 *al* 1988), and have very different vascular remanences, ranging from a few minutes to tens of  
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15 hours. An estimate of the time window during which a molecule of a given size may be  
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17 delivered across the BBB would be useful to monitor the amount of drug released to cerebral  
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19 tissue and to adjust the dosage. For example, it has been shown that 3 – 8 nm wide particles  
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21 are able to cross BBB 20 min after its ultrasound-induced disruption, but not particles of size  
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23 50 nm (Choi *et al* 2010). Additionally, as the brain is unprotected while the BBB is open, it is  
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25 important to know the time-to-recovery to limit any possible brain damages induced by  
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27 tissular penetration of pathogenic agents.  
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32 The quantification of the amount of MR-CA crossing the BBB requires development  
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34 of molecular imaging sequences via imaging techniques that are sensitive (high field MRI)  
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36 and provide outputs that can be correlated with the MR-CA concentration. Contrast agent  
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38 (CA) quantification is one of the principal challenges for MR molecular imaging. Unlike  
39  
40 other techniques such as fluorescence or nuclear imaging where the received signal comes  
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42 directly from the CA, MR signal comes from the surrounding water protons and is therefore  
43  
44 indirectly related to the presence of an exogenous probe. To quantify the CA concentration  
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46 using MRI, it is necessary to model its interaction with the surrounding water molecules.  
47  
48 Paramagnetic complexes interact with the surrounding water protons by decreasing their  
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50 relaxation times (Swift and Connick 1962). CA concentration can be linked to the  $T_1$  or  $T_2$   
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52 relaxation time decrease (in ms). In this study, we adapted a  $T_1$  mapping sequence proposed  
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54 by Deichmann and colleagues (Deichmann and Haase 1992; Deichmann *et al* 1999) that  
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permitted estimation of a wide range of  $T_1$  values with a good accuracy, and high spatial and temporal resolutions.

In this study, a calibrated BBB opening procedure was performed in healthy young adult rats under MR guidance. Five MR-CA were used (3 paramagnetic and 2 superparamagnetic) with different hydrodynamic diameters (from 1 to 65 nm) to investigate the maximum permissible size of the gaps induced in the endothelial wall under safe conditions. Then, using the  $T_1$  mapping strategy for the 3 paramagnetic CA we quantified the amount of particles crossing the BBB when injected at different times after ultrasound-induced BBB disruption. This enabled quantitative monitoring of the dynamics of BBB closure for each given molecular size. Lastly, we present a theoretical model to fit the experimental data and derive a calibration curve to predict duration of BBB opening as a function of the hydrodynamic diameter of a given CA.

## Materials and Methods

### Animal Preparation

All experiments were performed in accordance with the recommendations of the European Community (86/609/EEC) and the French legislation (decree no. 87/848) for use and care of laboratory animals. A total of 36 Sprague Dawley male rats (80-100 g, Janvier, France) were used. Their head was shaved to ensure proper coupling of the ultrasound beam to the brain. Rats were anesthetized with 1.5% isoflurane in a mixture of air and oxygen and placed in a cradle in prone position. A catheter (25 G needle) was positioned in the caudal vein to inject microbubbles and MR-CA from outside the scanner with minimal dead volume. 10% heparin was added to all injected solutions to avoid clot formation in the catheter. Body temperature and respiration rate were continuously monitored during the experiments.

## MR-CA

MR-CA of different hydrodynamic diameters were provided by Guerbet Research (France). Main characteristics and injected doses of these nanoparticles are summarized in Table 1. After each MR-CA injection, the bolus was flushed by injection of 100  $\mu$ l of saline solution. Three of the five MR-CA were paramagnetic Gd-chelates (Dotarem®, P846, and P792) and were detected and quantified using  $T_1$  strategies (Kang *et al* 2010; Protti *et al* 2010; Yankeelov *et al* 2006). The other two CA (P904 and P03680) were USPIO. A  $T_2$  -weighted sequence was used to detect them in cerebral tissues (Hyodo *et al* 2009; Philippens *et al* 2004). Longitudinal  $r_1$  and transverse  $r_2$  relaxivities were measured *in vitro* at 7T using galleries of tubes containing different concentrations of CA diluted in a 0.3% agar matrix, and maintained at 37°C.

## Ultrasound equipment

A MR-compatible focalized transducer (central frequency 1.5 MHz, diameter 30 mm, focal depth 20 mm, Imasonic, France) driven by a programmable function generator was used to produce ultrasound waves. It was coupled to the rat skull using a latex balloon filled with deionized and degassed water. Electrical power sent to the transducer was monitored during the BBB opening session. Main characteristics of the transducer (focal point size 0.6×0.6×3 mm<sup>3</sup>, transcranial acoustic transmission factor 52 ± 5 %) were estimated in a previous study [40].

## BBB opening

BBB disruption was performed during MR imaging session thanks to a dedicated holder maintaining the ultrasound transducer above rat head. Its position was monitored using the ARFI sequence and the right thalamus in the brain was chosen as a target for BBB



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3 disruption. Sonovue® microbubbles (Bracco, Italy) were administrated via a bolus ( $1.5 \times 10^8$   
4 bubbles/ml, 200  $\mu$ l, 2 s) approximately 5 s before the beginning of ultrasound session. The  
5 bolus was flushed by injection of 100  $\mu$ l of saline solution. Sonication was performed with  
6 3 ms bursts every 100 ms for one minute (Choi *et al* 2011). The peak negative acoustic  
7 pressure at the ultrasound focal point was calibrated to 0.45 MPa, a level sufficiently low to  
8 ensure safe and reversible BBB opening (Larrat *et al* 2011; Chopra *et al* 2010). Anatomical  
9 images were acquired at the end of each MRI session to verify integrity of the brain tissues.  
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### 23 **MRI acquisitions**

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25 MRI was performed on a 7T/90 mm Pharmascan scanner (Bruker, Germany). A saddle  
26 coil was specially designed in-house for excitation and signal reception. The geometry and  
27 size of the resonator (diameter 30 mm) were optimized to maximize the overall signal-to-  
28 noise ratio over the whole brain while allowing the ultrasound beam to propagate from the  
29 transducer to the rat brain.  
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36 A multislice spin-echo sequence was modified to include additional motion-sensitizing  
37 gradients, and synchronized to ultrasonic bursts so that the phase signal was proportional to  
38 local acoustic intensity (Larrat *et al* 2010; McDannold and Maier 2008). This ARFI sequence  
39 was acquired with the following parameters: TE/TR = 40/1700 ms,  $T_{acq} = 4$  min,  
40  $R = 0.5 \times 0.5 \times 1$  mm<sup>3</sup>, duration of the motion encoding gradients = 13.3 ms, duration of the  
41 sonication = 3 ms.  
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49 A high spatial resolution T<sub>2</sub>-weighted RARE sequence ( $TE_{eff}/TR = 32/4200$  ms,  $R =$   
50  $0.125 \times 0.125 \times 0.3$  mm<sup>3</sup>) was used to detect the presence of USPIOs, and also for acquiring  
51 images at the end of every experiment to confirm lack of haemorrhages or edema due to  
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3 ultrasound.  $T_1$ -weighted MSME sequence ( $TE/TR = 8/300$  ms,  $R = 0.250 \times 0.250 \times 1$  mm<sup>3</sup>) was  
4  
5 acquired to detect paramagnetic Gd-chelates.  
6

7 To measure concentration of Gd-chelates, a  $T_1$  mapping sequence was acquired before  
8 and after MR-CA injection. It consisted in a segmented series of fast gradient echo (FGE)  
9 images acquired at different time points after magnetization inversion in order to follow the  
10 entire  $T_1$  recovery curve [24, 25]. Sequence parameters were:  $TR_1$ , 5 ms, TE, 2.5 s, 6  
11 segments, 60 inversion times (from 64 to 5800 ms), flip angle, 5°, and  $R = 0.2 \times 0.2 \times 1$  mm<sup>3</sup>.  
12 Repetition time between the acquisitions of two segments,  $TR_2$ , was 9 s, and total acquisition  
13 time is 12.5 min. A centric encoding of the k-space was chosen in order to avoid  $T_2$  effects  
14 during echo trains at the acquisition of central lines.  
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### 29 **Data analysis**

30 Data were analyzed using dedicated codes written in Matlab software (MathWorks,  
31 USA). Maps of acoustic pressure index ( $IP_{ac}$ ) were deduced from ARFI sequence using the  
32 following equation:  
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$$38 \quad IP_{ac} \propto \frac{\sqrt{\varphi_{ON} - \varphi_{OFF}}}{P_{th}} \quad (1)$$

39 where  $\varphi_{ON}$  (respectively  $\varphi_{OFF}$ ) is the phase of ARFI signal acquired with ultrasound  
40 (respectively without ultrasound), and  $P_{th}$  is the expected acoustic pressure at the focal point  
41 in case of an ideal ultrasound coupling. In a previous study (Larrat *et al* 2010), we verified the  
42 linearity between acoustic intensity and MR-ARFI phase signal using the same setup. In  
43 another study (Larrat *et al* 2011), we also characterized and verified the linearity between  
44 acoustic pressure and the amount of MR-CA crossing the BBB in the range of acoustic  
45 pressure values applied here.  
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To generate  $T_1$  maps, the MRI signal measured with the FGE sequence was fitted as a function of the inversion time pixel by pixel as proposed by Deichmann and colleagues [24, 25]. MR-CA concentration maps ( $C$ ) were then calculated from the  $T_{10}$  and the  $T_1$  maps after MR-CA injection using the following equation (Swift and Connick 1962), considering that relaxivities  $r_1$  measured *in vitro* in agar matrix hardly differ from the ones in rat brain tissues:

$$C = \frac{1}{r_1} \left( \frac{1}{T_1} - \frac{1}{T_{10}} \right) \quad (2)$$

Signals from different manually drawn ROIs were analyzed. A  $ROI_{spot}$  of  $0.6 \times 0.6 \times 3 \text{ mm}^3$  corresponding to the transducer focal point size was drawn in the right thalamus. To analyze the effect of BBB disruption, a similar  $ROI_{contra}$  was taken in the corresponding contralateral (left thalamus) region of the brain. This enabled correction for the residual vascular concentration of MR-CA during imaging. Another control ROI was drawn in an extracerebral region in the cheek muscles ( $ROI_{muscle}$ ). For each rat, a corrected MR-CA concentration at the focal point ( $C^*$  without unit) was calculated as given below, based on the concentration measured at the  $ROI_{spot}$  ( $C_{ROI_{spot}}$ ) and that at the control spots ( $C_{ROI_{contra}}$  and  $C_{ROI_{muscle}}$ ), and the index of deposited acoustic pressure ( $IP_{ac}$ ):

$$C^* = \frac{C_{ROI_{spot}} - C_{ROI_{contra}}}{C_{ROI_{muscle}} \cdot IP_{ac}} \quad (3)$$

This processing step was mandatory to ensure the correction of the bias introduced by unequal injected volumes (unequal plasmatic concentrations) and the bias introduced by variations in acoustic wave penetration through the skull. These corrections allowed comparing MR-CA concentrations among animals.

## Results

### Low power, pulsed, ultrasound for localized disruption of BBB

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3 In a rat model, BBB was opened transiently with the use of ultrasound, immediately  
4 followed by intravenous injection of Dotarem® (gadolinium (Gd)-chelate) and measurement  
5 of the amount of the MR-CA delivered across the BBB (Figure 1). Before the ultrasound  
6 sonication, acoustic radiation force imaging (ARFI) was performed to obtain the acoustic  
7 pressure map in the brain (Figure 1-A). This sequence was used to verify that the ultrasound  
8 beam was focused at the desired point in the right thalamus, and to estimate the acoustic  
9 pressure at this focal point prior to microbubbles injection, ensuring that the acoustic pressure  
10 remains below a predetermined safety threshold (Chopra *et al* 2010; Larrat *et al* 2011). Once  
11 the transducer was set at the desired position, Sonovue® microbubbles were injected  
12 intravenously via a catheter positioned in the caudal vein, and the BBB was disrupted by  
13 pulsed sonication for one minute. Dotarem® was then injected using the same catheter. To  
14 measure the concentration of Dotarem® delivered, T<sub>1</sub> maps were acquired before and after  
15 MR-CA injection (Figure 1-B). There was high tissue perfusion of the CA (strong T<sub>1</sub>  
16 decrease) in the cheek and neck muscles, but not in the brain tissue due to the BBB. As  
17 expected, the ultrasound focal point (red arrow) was the only area of the brain exhibiting a  
18 strong T<sub>1</sub> decrease indicating BBB disruption. Based on T<sub>1</sub> mapping, CA concentration maps  
19 were obtained at the three selected regions of interest (ROIs): ultrasound focal point, ROI<sub>spot</sub>  
20 (n°1), a control region contralateral to the ultrasound focal point, ROI<sub>contra</sub> (n°2) and a region  
21 in cheek muscles ROI<sub>checks</sub> (n°3) (Figure 1-C).  
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#### 47 **Maximum gap between endothelial cells after reversible ultrasound-induced disruption** 48 **of the BBB** 49

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51 In order to determine the size of gaps generated after ultrasound-induced disruption of  
52 the BBB, we chose five different MR-CA of varying hydrodynamic diameters (Table 1) and  
53 verified their passage across BBB at the ultrasound focal point. The CA were intravenously  
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3 injected (n=1 rat for each CA) directly after ultrasound-induced BBB disruption. T<sub>1</sub>-weighted  
4 (T<sub>1w</sub>) images were obtained for the 3 paramagnetic CA and T<sub>2</sub>-weighted (T<sub>2w</sub>) images for the  
5 two ultrasmall superparamagnetic iron oxide (USPIO) CA (Figure 2). For all three Gd-  
6 chelates, a positive contrast was clearly visible on T<sub>1w</sub> images at the ultrasound focal point,  
7 revealing the presence of the CA. Student's *t*-test performed on the measured MR signals  
8 confirmed that signal differences between ROI<sub>spot</sub> and ROI<sub>contra</sub> were significant (p < 0.01).  
9 For the 25 nm USPIO, a negative contrast appeared on T<sub>2w</sub> images after its injection  
10 suggesting that BBB was permeable to molecules of this size. Signal differences with  
11 contralateral region were also significant (p < 0.01). However, the largest USPIO (65 nm) was  
12 detected only at the very focal point of the transducer as its penetration was significantly  
13 hindered compared to the smaller USPIO. The signal differences between ROI<sub>spot</sub> and  
14 ROI<sub>contra</sub> for this USPIO were not significant (p > 0.01) if the ROI was of the standard size  
15 (0.6×0.6×3 mm<sup>3</sup>), but the differences became significant (p < 0.01) if the size of the ROI<sub>spot</sub>  
16 was decreased to 0.2×0.2×1 mm<sup>3</sup> around the focal point. These results suggest that 65 nm is  
17 close to the maximal gap between endothelial cells that is attainable after BBB disruption  
18 with this protocol.  
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#### 41 **Quantification of MR-CA concentration**

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43 Rats were injected with Dotarem® at increasing doses (375, 750 and 1500 µmol/kg,  
44 n=1 rat per dose) following BBB disruption. Concentration maps acquired by T<sub>1</sub> mapping  
45 were overlaid on the anatomical T<sub>2w</sub> images (Figure 3-A). Additionally, a T<sub>1</sub>-weighted  
46 MSME sequence was also acquired. The estimated CA concentration at the ultrasound focal  
47 point was found to be proportional to the injected dose (C<sub>375</sub> < C<sub>750</sub> < C<sub>1500</sub>) (Figure 3-B). In  
48 contrast, the T<sub>1w</sub> signals did not exhibit the same behavior and even saturated at high CA  
49 injected doses (Figure 3-C). These observations confirm the high interest of our protocol to  
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3 precisely quantify small variations of CA concentration, and that quantitative analysis based  
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5 on  $T_{1w}$  signals alone can be misleading. Hence, results presented in the following sections of  
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7 this study were all derived from quantitative data obtained using the  $T_1$  mapping sequence  
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9 (Equation 3).  
10

### 14 **Quantitative study of BBB closure dynamics**

16 The BBB closure dynamics was assessed using paramagnetic CA of different  
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18 molecular sizes, administrated at different times after ultrasound-induced disruption. Rats  
19  
20 were injected with the following CA: Dotarem® (n=10 rats), P846 (n=10 rats) and P792 (n=8  
21  
22 rats) from 0 to 24 hours after ultrasound sonication. The reproducibility of CA delivery was  
23  
24 verified by measuring both the injected dose and the acoustic power deposited at the  
25  
26 ultrasound focal point. CA concentrations measured in the cheek muscles in each animal were  
27  
28 directly correlated with the injected dose since no permeability barrier like BBB exists in  
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30 muscles (Figure 4-A). Mean values were:  $0.345 \pm 0.034$  mM for Dotarem® group,  
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32  $0.061 \pm 0.008$  mM for P846 group and  $0.040 \pm 0.002$  mM for P792 group. Variabilities  
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34 observed within the above three groups (10%, 13% and 5%, respectively) could be explained  
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36 by experimental errors during injection and also by slight variations in the weight or  
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38 metabolism of the animals. An index of the acoustic pressure ( $IP_{ac}$ ) deposited at the focal  
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40 point was estimated from ARFI sequence (Figure 4-B) and was  $1.13 \pm 0.06$  for Dotarem®  
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42 group,  $1.00 \pm 0.05$  for P846 group and  $0.97 \pm 0.13$  for P792 group. Variabilities observed  
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44 within the above three groups (5%, 5% and 13%, respectively) was mainly explained by slight  
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46 differences in the placement of the transducer on the heads. Variations in the incidence angle  
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48 of the ultrasound beam or in the thickness of the skull could have a significant impact on  
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50 attenuation of the beam during its passage through the skull and lead to variations of the peak-  
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52 negative acoustic pressure at the focal point.  
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3 Corrected CA concentrations ( $C^*$ ) measured at the ultrasound focal point were plotted  
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5 as a function of the time elapsed between BBB disruption and MR-CA injection (Figure 4-C).  
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7 For Dotarem® (1 nm particle), the BBB was permeable for several hours post-BBB  
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9 disruption. A significant amount of CA was detectable in the brain even when injected 8 or 9  
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11 hours after disruption. Around 24 h post-disruption, the BBB had apparently recovered its  
12  
13 whole integrity, as very few CA were detectable in brain tissues. P846 (4 nm particle) was  
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15 delivered across the BBB only if injected within 2 hours post-BBB disruption. For P792  
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17 (7 nm particle), the duration of CA delivery to the brain was even shorter than for P846. One  
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19 hour post-BBB disruption, the amount of CA crossing the BBB was about 30% of its initial  
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21 value and dropped to undetectable levels after 90 minutes.  
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25 These findings confirmed that the duration of BBB crossing decreases when  
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27 increasing CA hydrodynamic diameter. Secondly, it appears from the shape of the time-  
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29 dependent concentration curve (Figure 4-C) that the BBB permeability decreased at a faster  
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31 rate initially and at a lesser rate later on.  
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### 34 35 36 **A model for BBB closure dynamics**

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38 Based on the experimental observations, the BBB closure dynamics was modelled as  
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40 follows. First, it was assumed that each individual gap generated in the BBB by ultrasound  
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42 returns to its equilibrium position (i.e. closed) as a harmonic oscillator damped by fluid  
43  
44 friction. The gap diameter as a function of the time was therefore expressed as  $d(t) = d_0 e^{-kt}$ ,  
45  
46 where  $d_0$  is the gap diameter just after BBB disruption and  $k$  a time constant representative of  
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48 individual gap closure speed.  
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52 Next, the initial size distribution of gaps was assumed to be a hemi-Gaussian function  
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54 centered on 0 with a standard deviation  $\sigma_0$ . These two conditions imply a distribution of gap  
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56 sizes as a function of time expressed as follows:  
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$$N(x,t) \propto \frac{1}{\sigma_0 e^{-kt}} e^{-\frac{x^2}{2\sigma_0^2 e^{-2kt}}} \quad (4)$$

For each time  $t$ , the amount  $Q_{CA}(t)$  of CA of hydrodynamic diameter  $d_H$  crossing the BBB in a unit volume post-disruption is proportional to the inflow flux only, since the outflow of CA from brain to blood is assumed to be negligible due to high concentration in the blood compartment. Assuming a hard sphere behavior for CA molecules,  $Q_{CA}(t)$  can be expressed as follows:

$$Q_{CA}(t) \propto \int_{d_H}^{+\infty} N(x,t) \cdot \frac{x^2}{d_H} dx \quad (5)$$

The CA diffusion in brain outside of the BBB disrupted region is also neglected since the CA tissular concentration is quantified within a few minutes after CA intravenous injection. A previous study from our team (Marty *et al* 2010) showed that the Gd-chelates used in this study do not diffuse fast enough in brain parenchyma to make this effect significant: for example, the diffusion coefficient of Dotarem® (1 nm particle) was estimated around  $46 \mu\text{m}^2 \cdot \text{s}^{-1}$  (Table 1).

Integration of equation 5 gives:

$$Q_{CA}(t) \propto \frac{\sigma_0^2 e^{-2kt}}{d_H} \left( \sqrt{\frac{\pi}{2}} \left( 1 - \text{erf} \left( \frac{d_H}{\sqrt{2}\sigma_0 e^{-kt}} \right) \right) + \frac{d_H}{\sigma_0 e^{-kt}} e^{-\frac{d_H^2}{2\sigma_0^2 e^{-2kt}}} \right) \quad (6)$$

All experimental data (n=28 rats) were used to fit this theoretical BBB closure model and determine the two parameters: a characteristic gap diameter  $\sigma_0 = 1.54 \text{ nm}$ , and BBB closure rate  $k = 1.54 \times 10^{-5} \text{ s}^{-1}$ .

The half closure time ( $t_{1/2}$ ) was then defined as the time after disruption when the concentration of CA crossing the BBB was 50% of the maximal concentration obtained immediately after disruption.  $t_{1/2}$  was then calculated as the numerical solution of the following equation:



$$\frac{1}{2} \left( \sqrt{\frac{\pi}{2}} \left( 1 - \operatorname{erf} \left( \frac{d_H}{\sqrt{2}\sigma_0} \right) \right) + \frac{d_H}{\sigma_0} e^{-\frac{d_H^2}{2\sigma_0^2}} \right) = e^{-2kt_{1/2}} \left( \sqrt{\frac{\pi}{2}} \left( 1 - \operatorname{erf} \left( \frac{d_H}{\sqrt{2}\sigma_0 e^{-kt_{1/2}}} \right) \right) + \frac{d_H}{\sigma_0 e^{-kt_{1/2}}} e^{-\frac{d_H^2}{2\sigma_0^2 e^{-2kt_{1/2}}}} \right) \quad (7)$$

From the fitted parameters ( $\sigma_0$  and  $k$ ), half closure times could be estimated as a function of hydrodynamic diameters of the molecules by numerically solving equation 7 (Figure 5, black circles). The calculated half closure time was  $\sim 5.5$  h for a 1 nm particle;  $\sim 1.5$  h for a 4 nm particle and about 30 min for a 7 nm particle. These values were close to those estimated experimentally for Dotarem®, P846 and P792, which were 5.2 h, 1.2 h and 45 min, respectively (Figure 5, red crosses).

For larger molecules, half closure times were greatly reduced (around 3 min for a 25 nm particle and only few seconds for a 65 nm particle). These results match the qualitative results obtained with USPIOs (Figure 2). Therefore, to deliver a significant amount of larger molecular size CA across the BBB their intravenous circulation should coincide with the end of ultrasound sonication.

Lastly, it appears that the following analytical formula fits reasonably well the numerical solution of equation 7, enabling to directly estimate half closure time as a function of hydrodynamic diameter:

$$t_{1/2} = \frac{A}{1 + B \cdot d_H^2} \quad (8)$$

with  $A = 2.35 \times 10^4$  s and  $B = 0.2106 \text{ nm}^{-2}$  (Figure 5, solid line).

## Discussion

### Ultrasound-induced BBB disruption

Drug delivery to the brain remains a major challenge for the treatment of most brain diseases. Although pathologies such as cancer degrade BBB integrity (Liu *et al* 2012), it was proven that endothelial wall permeability is still poorer in these regions than in vessels of

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2  
3 other organs. Recently, ultrasound-induced BBB disruption was demonstrated to be an  
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5 effective method of significantly increasing permeability of the blood-tumor barrier in a  
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7 controlled manner (Chen *et al* 2010). Another study reported that vessel wall permeability  
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9 after ultrasound-induced BBB disruption was comparable to permeability values for  
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11 unprotected organs (Vlachos *et al* 2010). Our study further demonstrates that ultrasound  
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13 combined with intravenous injection of microbubbles is a reliable, reproducible and non-  
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15 invasive technique to deliver nanoparticles with hydrodynamic diameters up to 65 nm.  
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17 Moreover, in our protocol, no edema or hemorrhages were detected on  $T_{2w}$  images following  
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19 ultrasound sonication. Animals kept alive for several weeks after undergoing BBB disruption  
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21 showed no adverse effects. The observation of a return to basal permeability within 24 hours  
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23 for the smallest available CA (1 nm) further proved that the disruption was transient and  
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25 reversible.  
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### 32 **MR guidance**

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34 The ultrasound-induced BBB disruption described here was performed entirely under  
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36 high-field MRI guidance. This is a challenge as it requires integrating a stereotactic frame, a  
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38 MR-compatible ultrasound transducer adapted to the geometry of the rodent head and a  
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40 dedicated radiofrequency (RF) coil fitted into the 8.5 cm bore of a preclinical MRI scanner.  
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42 The optimized RF coil allowed the quantification of Gd MR-CA at a micromolar range, with  
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44 sufficient signal homogeneity over the whole brain and the ability for the ultrasound beam to  
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46 propagate through the coil.  
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50 The anesthetized animal was placed inside the MRI scanner just before the start of the  
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52 BBB disruption procedure till recovery. Thus, it was possible to obtain an acoustic intensity  
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54 map before BBB disruption. This ensured a proper positioning of the ultrasound focal spot in  
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56 the right thalamus and a precise *in situ* calibration of the acoustic pressure to avoid  
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3 irreversible tissue damage. Furthermore, microbubbles were injected to cause ultrasound-  
4 induced BBB disruption followed by injection of MR-CA via the same the catheter, which  
5 allowed imaging of the early events post-BBB disruption.  
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### 9 10 11 **Maximum gap obtained between endothelial cells**

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14 CA of increasing hydrodynamic sizes were tested, and the maximum gap width  
15 generated between endothelial cells after BBB disruption was estimated with our acoustic  
16 parameters. Gaps slightly above 65 nm could be generated safely, and the BBB recovered full  
17 integrity within 24 h. The mechanical index of our set-up was 0.37 which is below the  
18 standard value proposed by other studies for reversible BBB opening (McDannold *et al*  
19 2008). This result is particularly interesting from the view of development of targeted CA.  
20 Using a similar sonoporation protocol, it would be possible to deliver nanoparticles such as  
21 USPIOs, functionalized to target tissular biomarkers. In contrast, larger objects like emulsions  
22 or liposomes that typically have hydrodynamic diameters >100 nm, would be difficult to  
23 deliver to the brain using this technique, assuming a purely paracellular pathway. In general,  
24 an estimate of the maximum possible gap diameter would be helpful when the technique is  
25 applied to deliver therapeutic molecules. Several therapeutic drugs could benefit from this  
26 information: for instance chemotherapeutic drugs such as Avastin® ( $d_H \sim 10$  nm),  
27 recombinant adeno-associated virus (rAAV) vectors for gene therapy ( $d_H \sim 20$  nm),  
28 monoclonal antibodies such as anti- $\beta$ -amyloid ( $d_H$  of few nm), and vectorized short interfering  
29 RNA (siRNA) used for gene expression modulation ( $d_H \sim 50$  nm).  
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50 The maximum gap diameter measured in this study is probably dependent of the  
51 sonication parameters, in particular ultrasound frequency, pulse duration and duty cycle, total  
52 sonication time and size distribution of injected microbubbles (Sonovue® has a broad  
53 distribution between 1-8  $\mu$ m). Here, we chose well established parameters that have been  
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3 proposed by others (Choi *et al* 2011). Several groups have studied the influence of acoustic  
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5 parameters (O'Reilly *et al* 2011a) and microbubbles size (Samiotaki 2011) on BBB opening.  
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7 An optimization of parameters based on the knowledge gained from these studies may allow  
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9 generation of pores larger than 65 nm wide.  
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### 11 12 13 14 **BBB closure dynamics**

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16 In this study, the concentration of paramagnetic CA delivered at the ultrasound focal  
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18 point was quantitatively measured, based on which, the duration of passage across the BBB  
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20 was estimated for molecules of different sizes. This duration decreased rapidly as a function  
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22 of hydrodynamic diameter of the CA. For instance, small molecules (around 1 nm) continued  
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24 to cross the BBB for more than 10 h, whereas large iron oxide particles (around 25 nm) had a  
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26 time window of only a few minutes. Therefore, to maximize the amount of extravasation in  
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28 the brain, large molecules should be injected right after or even during ultrasound sonication.  
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32 The results presented here also suggest a closure of the BBB at a progressively  
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34 decreasing rate. We present a simple, damped, elastic model assuming an exponential decay  
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36 of individual gap sizes to semi-quantitatively describe the observed dynamics of BBB closure.  
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38 The experimental data were fed in the model to derive important physiological parameters,  
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40 such as the typical closure time constant  $k$  that was estimated to be  $1.54 \times 10^{-5} \text{ s}^{-1}$ . This  
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42 parameter is an indirect measurement of rate of endothelial cell relaxation after contraction. It  
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44 could be used as a physio-pathological index of vessel integrity.  
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48 This model assumes a paracellular passage of injected molecules with an inflow  
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50 limited by the size of the molecule relative to the size of the pores. Our results support the  
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52 notion that most of the crossing happens between endothelial cells rather than through them.  
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54 This reinforces the hypothesis that during ultrasound-induced BBB opening sonicated  
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3 microbubbles exert a mechanical stress on endothelial cells which makes them contract on  
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5 themselves and loosen the junctions between them creating gaps (Vykhodtseva *et al* 2008).  
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### 8 9 **Limitations of the study**

10 The proposed BBB closure model has certain limitations. It does not take into account  
11 potential increased permeability of endothelial cell membranes (transcellular BBB crossing).  
12 It does not distinguish between hydrophilic and lipophilic molecules although it is known that  
13 lipophilic agents present an increased ability to cross the BBB via transcytosis (Pardridge  
14 2005). This concurrent release pathway could explain the observed differences between  
15 experimental data of P846 and P792 and predictions from the theoretical model.  
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18 The three Gd-chelates compared in this study have similar but not equal vascular  
19 remanences (Table 1). The tissular concentrations were measured within 12.5 minutes after  
20 CA injection, which is not a negligible time delay as the plasmatic half-life of the CA are  
21 between 26 and 51 min (Fries *et al* 2009). Although during data processing the differences in  
22 overall injected plasmatic concentration were corrected for each CA (Equation 3), temporal  
23 variations of vascular concentrations during the T<sub>1</sub> mapping acquisition could be different  
24 from one CA to the other, none of them remaining strictly steady. The impact of these  
25 variations may be low since the MRI mapping acquisition was performed with a centric  
26 encoding scheme. Nevertheless, it may explain the differences between the theoretical model  
27 and the experimentally measured closure dynamics for the different molecules.  
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### 49 **Conclusion**

50 In this study, we developed a complete methodology for noninvasive, reversible and  
51 controlled ultrasound-induced BBB disruption under MR-guidance that permitted precise  
52 quantification of the amount of MR-CA delivered to brain parenchyma in a rat model. The  
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3 two key parameters in designing drug delivery across the BBB, namely, the largest  
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5 deliverable molecular size of the agent, and the time window for delivery of an agent of a  
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7 given molecular size, were estimated for the first time.  
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15  
16 RF coil support and Dr. A. Alahari for editing this manuscript.  
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### 19 20 21 **Disclosure/Conflict of interest**

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23 Authors declare no conflict of interest.  
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## Titles and legends to figures

**Figure 1: Different steps of MRI-guided ultrasound-induced disruption of blood-brain barrier (BBB).** BBB disruption was achieved by intravenous injection of microbubbles via a catheter in the tail vein, in conjunction with transcranial ultrasound sonication. This was followed by injection of MRI contrast agent (CA) via the same catheter. The entire procedure was performed inside a 7T MRI scanner. A. Index of deposited acoustic pressure map derived from ARFI images acquired before injection of microbubbles revealed the ultrasound focal point. B.  $T_1$  maps (in ms) acquired before (Baseline) and after MR-CA injection. C. MR-CA concentration map (in mM) derived from corresponding  $T_1$  maps. Concentrations from three regions of interest (ROIs) were analyzed: ultrasound focal point ( $n^{\circ}1$ ), contralateral control region ( $n^{\circ}2$ ) and extra-cerebral control region in the cheek muscles ( $n^{\circ}3$ ).

**Figure 2: Ultrasound-induced localized delivery in rat brain of MRI contrast agents (CA) with different hydrodynamic diameters.** Paramagnetic CA (Dotarem®, P846 and P792) or ultrasmall superparamagnetic iron oxide (USPIO) CA (P904 and P03680) were injected in rats immediately after ultrasound-induced BBB disruption, and the signal intensities were measured within 5 minutes at the ultrasound focal point.  $T_{1w}$  images of the paramagnetic CA (A, B and C) revealed a positive contrast, while  $T_{2w}$  images of the USPIO revealed a negative contrast (D and E). Comparison of the signal enhancement in arbitrary units (a.u.) in the region of interest of ultrasound focal point ( $ROI_{spot}$ ,  $n^{\circ}1$ ) and that in the control contralateral region ( $ROI_{contra}$ ,  $n^{\circ}2$ ) is given next to the scan corresponding to each CA ( $n = 1$  rat for each CA). Differences between the two regions were considered significant for  $p$ -values  $< 0.01$  (\*) using a Student's  $t$ -test.

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3 **Figure 3: Comparison of  $T_{1w}$ -weighted ( $T_{1w}$ ) and  $T_1$  mapping strategies to assess the**  
4 **amount of Gadolinium (Gd)-chelate crossing the BBB.** Dotarem® was injected at different  
5 doses just after ultrasound-induced BBB disruption in the right thalamus, followed by the  
6 acquisition of one  $T_1$  mapping sequence (to produce a concentration map) and one  $T_{1w}$   
7 sequence. A. Overlay of CA concentration maps on the corresponding anatomical  $T_{2w}$  images  
8 after injection of increasing doses of Dotarem® (375 , 750 and 1500  $\mu\text{mol/kg}$ ,  $n = 1$  rat per  
9 dose). Purple arrows indicate the ultrasound focal spot where the BBB was disrupted. B. CA  
10 concentration as estimated using  $T_1$  mapping strategy at the ultrasound focal point showing  
11 linear dependence on the injected dose. C.  $T_{1w}$  signal of the CA at the ultrasound focal point  
12 as a function of the injected dose showing a saturation effect.  
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27 **Figure 4: BBB closure dynamics after ultrasound induced disruption for MR-CA of**  
28 **different molecular sizes.** Paramagnetic CA were injected at different times (0 to 24h) after  
29 BBB disruption. A. Concentrations of the three paramagnetic CA in the region of interest in  
30 the cheek muscles  $C_{ROI\text{muscle}}$  were measured. B. Index of delivered acoustic pressure at the  
31 ultrasound focal point in the right thalamus in the brain. The maximum variability within each  
32 group (CA) for these two parameters was 13 %. C. Corrected concentration ( $C^*$ ) of the CA at  
33 the ultrasound focal point in the brain was plotted as a function of the time elapsed between  
34 ultrasound-induced BBB opening and the MR-CA injection ( $n = 1$  rat per time point). All  
35 experimental data ( $n=28$  rats) were used to fit the theoretical BBB closure model and  
36 determine the two parameters ( $\sigma_0 = 1.54$  nm,  $k = 1.54 \times 10^{-5}$  s $^{-1}$ ) (Equation 6); dashed lines  
37 correspond to the fitted data for each MR-CA.  
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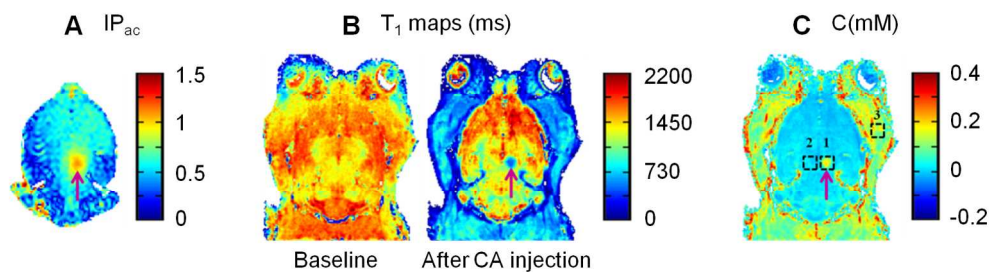
55 **Figure 5: Prediction of the time window of BBB passage for a given nanoparticle size.**

56 Based on the theoretical model proposed here, half closure time,  $t_{1/2}$  (the time after  
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3 ultrasound-induced BBB disruption that is required for the delivery of 50% of the maximal  
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5 dose to the predetermined ultrasound focal point in the brain) was calculated as a function of  
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7 the hydrodynamic diameter of the nanoparticle (Equation 7) (black circles). Red crosses  
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9 correspond to experimental data on the paramagnetic nanoparticle contrast agents used here  
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11 (Dotarem® 1 nm; P846 4 nm; and P792 7 nm) and the solid line represents the best fit with  
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13 analytical function:  $t_{1/2} = A / (1 + B \cdot d_H^2)$ , where  $d_H$  is the hydrodynamic diameter of the  
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15 injected CA (in nm) and A and B are constants,  $A = 2.35 \times 10^4$  s and  $B = 0.2106 \text{ nm}^{-2}$ .  
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	$d_H$ (nm)	$r_1$ ( $mM^{-1}.s^{-1}$ )	$r_2$ ( $mM^{-1}.s^{-1}$ )	Dose ( $\mu mol/kg$ body weight)	Vascular remenance (min)	Diffusion coefficient ( $\times$ $10^{-11} m^2.s^{-1}$ )
<b>Paramagnetic MR-CA</b>						
Dotarem®	<1	3.4	4.8	1500	26	4.6
P846	4	10.3	37	95	51	1.4
P792	7	4.9	42	210	32	0.8
<b>Superparamagnetic MR-CA</b>						
P904	21	1.4	88	1030	-	-
P03680	65	2.0	80	258	-	-

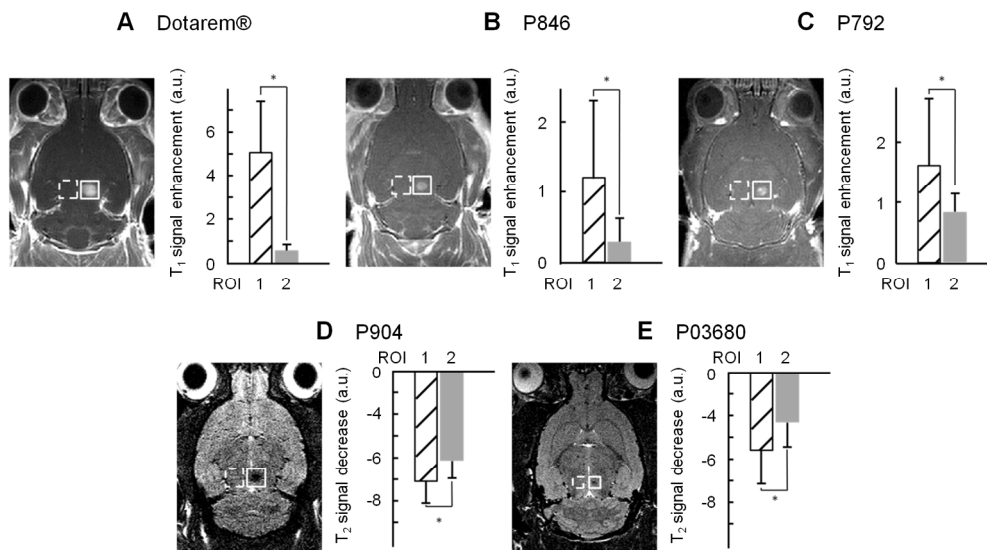
**Table 1: Principal features of the paramagnetic and superparamagnetic nanoparticle contrast agents (CA) used in this study.** Hydrodynamic diameters ( $d_H$ ) were measured by light scattering, longitudinal ( $r_1$ ) and transverse ( $r_2$ ) relaxivities were estimated at 7T in 0.3% agar gels maintained at 37°C, vascular remanences were taken from literature (34) and diffusion coefficients in brain parenchyma were estimated in a previous work (28). The paramagnetic CA were imaged using  $T_{1w}/T_1$  mapping sequences, while the superparamagnetic using  $T_{2w}$  sequences.



Different steps of MRI-guided ultrasound-induced disruption of blood-brain barrier (BBB). BBB disruption was achieved by intravenous injection of microbubbles via a catheter in the tail vein, in conjunction with transcranial ultrasound sonication. This was followed by injection of MRI contrast agent (CA) via the same catheter. The entire procedure was performed inside a 7T MRI scanner. A. Index of deposited acoustic pressure map derived from ARFI images acquired before injection of microbubbles revealed the ultrasound focal point. B.  $T_1$  maps (in ms) acquired before (Baseline) and after MR-CA injection. C. MR-CA concentration map (in mM) derived from corresponding  $T_1$  maps. Concentrations from three regions of interest (ROIs) were analyzed: ultrasound focal point ( $n^{\circ}1$ ), contralateral control region ( $n^{\circ}2$ ) and extra-cerebral control region in the cheek muscles ( $n^{\circ}3$ ).

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Ultrasound-induced localized delivery in rat brain of MRI contrast agents (CA) with different hydrodynamic diameters. Paramagnetic CA (Dotarem®, P846 and P792) or ultrasmall superparamagnetic iron oxide (USPIO) CA (P904 and P03680) were injected in rats immediately after ultrasound-induced BBB disruption, and the signal intensities were measured within 5 minutes at the ultrasound focal point.  $T_{1w}$  images of the paramagnetic CA (A, B and C) revealed a positive contrast, while  $T_{2w}$  images of the USPIO revealed a negative contrast (D and E). Comparison of the signal enhancement in arbitrary units (a.u.) in the region of interest of ultrasound focal point ( $ROI_{spot}$ ,  $n^{\circ}1$ ) and that in the control contralateral region ( $ROI_{contra}$ ,  $n^{\circ}2$ ) is given next to the scan corresponding to each CA ( $n = 1$  rat for each CA). Differences between the two regions were considered significant for p-values  $< 0.01$  (\*) using a Student's  $t$ -test.

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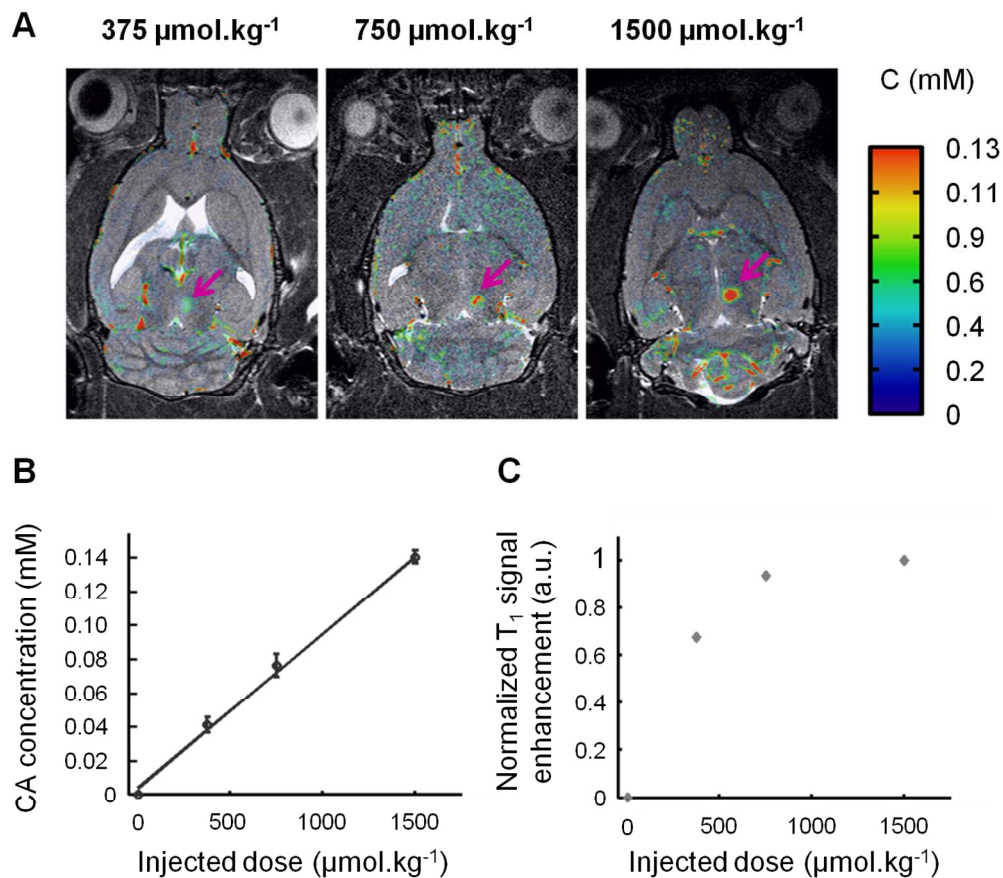


Figure 3: Comparison of  $T_1$ -weighted ( $T_{1w}$ ) and  $T_1$  mapping strategies to assess the amount of Gadolinium (Gd)-chelate crossing the BBB. Dotarem® was injected at different doses just after ultrasound-induced BBB disruption in the right thalamus, followed by the acquisition of one  $T_1$  mapping sequence (to produce a concentration map) and one  $T_{1w}$  sequence. A. Overlay of CA concentration maps on the corresponding anatomical  $T_{2w}$  images after injection of increasing doses of Dotarem® (375, 750 and 1500  $\mu\text{mol/kg}$ ,  $n = 1$  rat per dose). Purple arrows indicate the ultrasound focal spot where the BBB was disrupted. B. CA concentration as estimated using  $T_1$  mapping strategy at the ultrasound focal point showing linear dependence on the injected dose. C.  $T_{1w}$  signal of the CA at the ultrasound focal point as a function of the injected dose showing a saturation effect.

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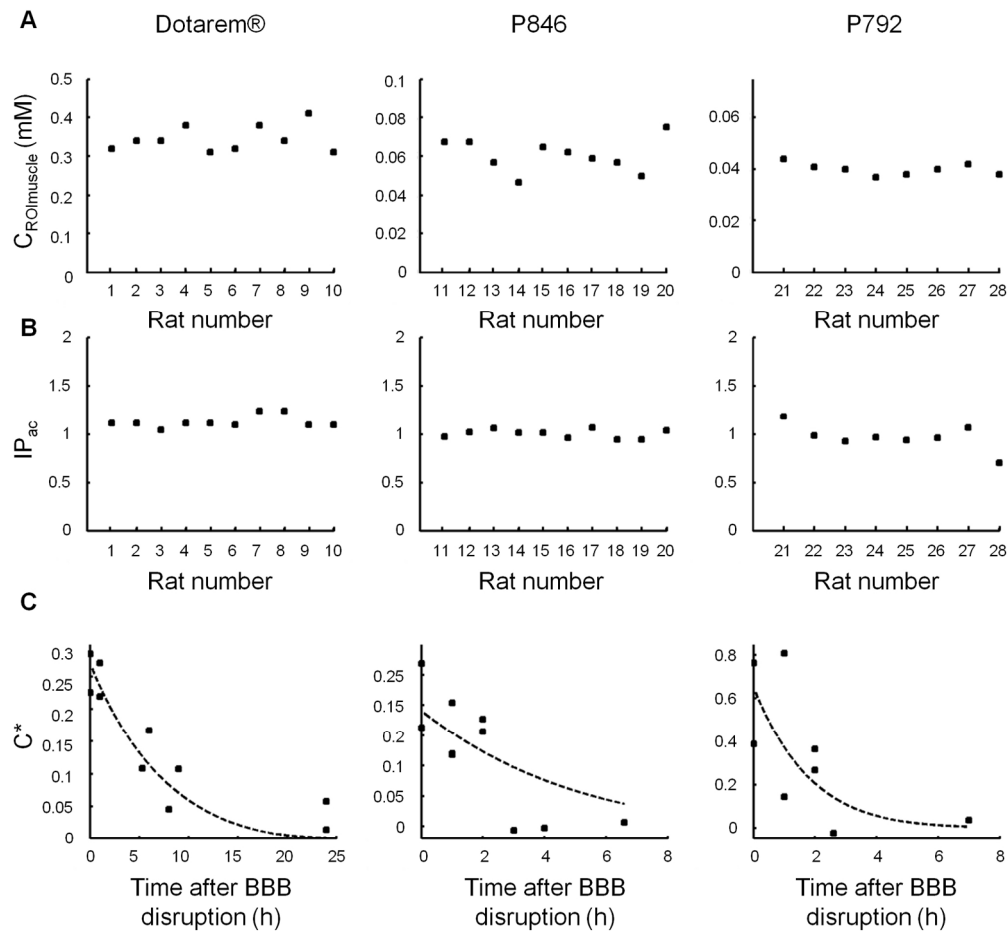
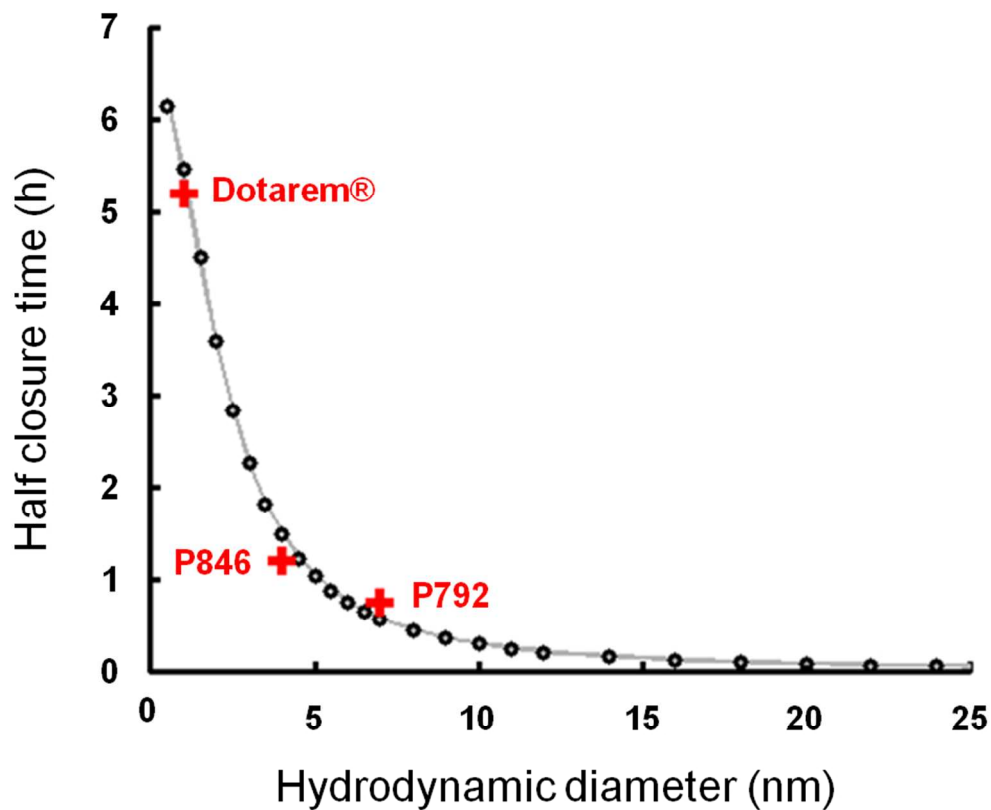


Figure 4: BBB closure dynamics after ultrasound induced disruption for MR-CA of different molecular sizes. Paramagnetic CA were injected at different times (0 to 24h) after BBB disruption. A. Concentrations of the three paramagnetic CA in the region of interest in the cheek muscles  $C_{ROImuscle}$  were measured. B. Index of delivered acoustic pressure at the ultrasound focal point in the right thalamus in the brain. The maximum variability within each group (CA) for these two parameters was 13 %. C. Corrected concentration ( $C^*$ ) of the CA at the ultrasound focal point in the brain was plotted as a function of the time elapsed between ultrasound-induced BBB opening and the MR-CA injection ( $n = 1$  rat per time point). All experimental data ( $n = 28$  rats) were used to fit the theoretical BBB closure model and determine the two parameters ( $\sigma_0 = 1.54 \text{ nm}$ ,  $k = 1.54 \times 10^{-5} \text{ s}^{-1}$ ) (Equation 6); dashed lines correspond to the fitted data for each MR-CA.

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Prediction of the time window of BBB passage for a given nanoparticle size. Based on the theoretical model proposed here, half closure time,  $t_{1/2}$  (the time after ultrasound-induced BBB disruption that is required for the delivery of 50 % of the maximal dose to the predetermined ultrasound focal point in the brain) was calculated as a function of the hydrodynamic diameter of the nanoparticle (Equation 7) (black circles). Red crosses correspond to experimental data on the paramagnetic nanoparticle contrast agents used here (Dotarem® 1 nm; P846 4 nm; and P792 7 nm) and the solid line represents the best fit with analytical function:  $t_{1/2} = A / (1 + B \cdot d_H^2)$ , where  $d_H$  is the hydrodynamic diameter of the injected CA (in nm) and A and B are constants,  $A = 2.35 \times 10^4$  s and  $B = 0.2106 \text{ nm}^{-2}$ .

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