Investigation of the maximum local specific absorption rate in 7 T magnetic resonance with respect to load size by the use of electromagnetic simulations

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Short running title: Maximum SAR in 7 T MR with respect to load size

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Abstract- The evaluation of the local specific absorption rate (SAR) is a major concern in ultra high field (UHF) magnetic resonance (MR) systems. In fact, at UHF, radiofrequency (RF) field inhomogeneity generates hot-spots that could cause localized tissue heating. Unfortunately, local SAR measurements are not available in present MR systems; thus, electromagnetic simulations must be performed for RF fields and SAR analysis. In this study we use three-dimensional (3D) full-wave numerical electromagnetic simulations to investigate the dependence of the local SAR at 7.0 T with respect to the subject size in two different scenarios: surface coil loaded by adult and child calves and quadrature volume coil loaded by adult and child heads. In the surface coil scenario, the maximum local SAR decreases with decreasing load size, provided that the RF magnetic fields for the different load sizes are scaled to achieve the same slice average value. On the contrary, in the volume coil scenario, maximum local SAR was up to 15% higher in children than in adults.

Keywords: ultra high field Magnetic Resonance, SAR, RF full-wave simulations

INTRODUCTION

There has been a long debate on possible differences in specific absorption rate (SAR) with respect to the load size. This debate has involved the SAR due to radiofrequency (RF) plane wave [Conil et al., 2008; Piuzzi et al., 2011] or to a cell phone [Christ et al., 2005], while such debate in magnetic resonance (MR) seems to be missing. Concerning the SAR due to a plane wave, the results presented in Conil et al. [2008] and Piuzzi et al. [2011] showed a trend towards higher SAR when reducing the dimension of the load (i.e., in children), while the conclusions of Christ et al. [2005] do not confirm the same trend. In MR, scanners give an estimation of the average SAR in the sample under test during the exam; in fact, during MR exams, average SAR exposure has to be monitored to remain below the regulatory limit of 3.2 W/kg imposed by the International Electrotechnical Commission (IEC), as established in regulation number IEC-60601-2-33. This SAR measure is obtained by means of an empirical formulation, which takes into account patient parameters (such as body size, gender, mass) and the MR sequence being used. However, the average SAR does not take into account the spatial inhomogeneity in the applied RF field, which can generate hot spots. It follows that a major concern is the correct evaluation of the local SAR, especially in ultra highfield (UHF) MR systems ($\geq$7.0T) [van Osch et al., 2014]. In fact, at UHF, the energy deposition due to the RF field increases and its distribution inside the subject becomes extremely inhomogeneous and subject-dependent [Collins et al., 1998; Collins, 2009]. The increased RF energy deposition and spatial variability at UHF is due to the higher operational frequency of the UHF MR system, which is equal to 298 MHz at 7 T [Vaughan et al., 2001; Collins et al., 2011]. Local SAR measurements are not available in current MR systems; thus, electromagnetic simulations must be performed for the analysis of RF fields and SAR [Collins et al., 1998 and 2004; Kozlov et al., 2009; van Lier et al., 2012; de Greef et al., 2013; Wolf et al., 2013]. Specifically, electromagnetic simulations permit the calculation of interactions between coils (i.e., the RF sources) and loads (i.e., anatomic human models comprising different tissues, each one characterized by its own dielectric properties). In Collins et al. [1998] local SAR was simulated in one single human head model for different static magnetic field strength, including 7 T; in Collins et al. [2004] and van Lier et al. [2012] authors have also addressed the relation between SAR and temperature using Pennes' bioheat equation. In Kozlov et al. [2009] the local SAR at 7 T MR was calculated for two different loads (one adult and one child), while in de Greef et al. [2013] the authors evaluated the local SAR at 7 T MR in four adults and two children with the aim of computing a safety factor accounting for inter-subject variability. In Wolf et al. [2013], two adult models (one male and one female) have been analyzed to determine a reliable SAR prediction for head with simplified tissue structures and model extent.

SAR calculation relies on simulated data from human models, but such models cannot match precisely the anatomy of actual subjects. Subject-specific anatomic simulations models would require to previously acquire and segment images of each subject [Homannet al., 2011; Voigh et al., 2012]: however, such approach is rarely feasible and therefore models are used instead. This common practice introduces a mismatch between real and simulated data, which can be compensated by simulating and comparing different human models [Collins, 2009; de Greef et al., 2013]. Here, we use three-dimensional (3D) full-wave numerical electromagnetic simulations to investigate the dependence of the maximum local SAR at 7 T with respect to the load size in two different scenarios: surface coil loaded by adult and child calves and quadrature volume coil loaded by adult and child heads. It will be shown that, in the first scenario, the maximum local SAR decreases in children: this holds true if the RF magnetic fields (denoted in MR community as B_1^+) for the different load sizes are scaled to achieve the same B_1^+ slice average value; on the contrary, in volume coil scenario, the maximum local SAR in children can be greater than that in adults.

METHODS

For the 3D full-wave numerical electromagnetic simulations, we used the finite integration technique (FIT) in time-domain employed in the Computer Simulation Technology Microwave Studio (CST MWS) Suite (CST-Computer Simulation Technology AG, Darmstadt, Germany). For the first scenario, we simulated a 1H single loop of radius 5 cm positioned near the human calf extracted from the 1×1×1 mm^3 voxel-size anatomic adult human model HUGO (CST MWS Suite), as shown in Figure 1a. Specifically, the simulated loop was constituted of copper and it had external radius of 5 cm, width of 0.5 cm, thickness of 35 μm. The loop was tuned to a frequency of 298 MHz (the operational RF for 7 T MR systems) and matched, achieving $S_{11} = -10$ dB. Details of the tuning and matching capacitor can be found in Stara et al. [2013], together with the surface coil model validation. RF fields and SAR inside the calf were calculated when applying 1W of input power. The average B_1^+ magnitude calculated in the axial slice crossing the coil center was computed, together with maximum of the local SAR (10 g, continuous RF wave excitation) in the same slice. For these simulations, approximately 4 million mesh nodes have been used (simulation time=16 h on one workstation). Next, we repeated the simulations using other human calves extracted from 2 anatomic adult human models (Virtual Population, IT'IS foundation, Zurich, Switzerland) and 4 anatomic child models (Virtual Population, IT'IS foundation, Zurich, Switzerland) [Christ et al., 2010; Hasgall et al., 2013]. In all the simulations, particular care was taken to keep the distance between the calves and the center of the coil fixed to 15 mm.

An overview of the details of anatomic models and of the segmented tissue types is given in Table 1 and Table 2, respectively.

In the second scenario, we used the same FIT to simulate a volume coil loaded by a human head. Specifically, we simulated a transmit-receive shielded 16 elements 1H high-pass birdcage head coil manufactured by Nova Medical (Wilmington, MA, USA), operating in quadrature at 298 MHz. The elements (copper flat strips having width of 2.5 cm and thickness of 35 μm) of the coil are placed equally spaced along a circle of radius of 29.5 cm; the elements are connected through two copper end-rings (having width of 1 cm). The radius of the copper shield is 37.5 cm while the length is 27.5 cm. The coil was loaded by a human head extracted from the 2×2×2 mm^3 voxel-size anatomic adult human model Ella (Virtual Population, IT'IS foundation) as shown in Figure 2a. Quadrature feeding has been employed using 4 sources, equally displaced by $\pi/2$ azimuthally, with a relative electrical phase shift of $\pi/2$. The coil has been tuned to the frequency of 298 MHz (the operational RF for 7 T MR systems) and matched using a capacitive matching circuit [Mispelter et al., 2006], achieving Sxx<-10 dB and Sxy<-10 dB. RF fields and SAR inside the head were calculated when applying 1 W of input power at each source. The average B_1^+ magnitude calculated in the axial slice crossing the eyes was computed, together with maximum local SAR (10 g, continuous RF wave excitation) calculated in the entire head. For these simulations, approximately 12.5 million mesh nodes have been used (simulation time=15 h on one workstation using the Nvidia Quadro 6000 (Nvidia, Santa Clara, CA, USA) graphics processing unit (GPU) computing). Afterwards, we repeated the simulations using other human heads extracted from the anatomic models listed in Table 1.

To validate the birdcage coil model, we performed a comparison between measured and simulated B_1^+ magnitude: more in detail, we measured the B_1^+ magnitude of a 0.1 M saline solution cylindrical phantom (radius=6 cm, height=24 cm), positioned in the coil as shown in Figure 3a, by using the Bloch-Siegert Shift (BSS) method [Sacolick et al., 2010] on a GE MR950 7 T human system (GE HealthCare, Milwaukee, WI, USA) equipped with the birdcage coil described above. In the simulation, the 0.1 M saline solution load was modeled as a lossy homogeneous dielectric cylinder having permittivity of 80 and conductivity of 0.6 S/m, as derived from Tiberi et al. [2013].

RESULTS

Figure 1b, c shows the maps of B_1^+ magnitude [μT] and SAR [W/Kg] in the axial slice crossing the coil center obtained by using CST MW Suite with the human calf extracted from HUGO. Both maps, which refer to continuous RF wave excitation, are given only in the region occupied by the load.

Table 3 summarizes the results obtained by applying CST MW Suite to the human calves extracted from all the models listed in Table 1. Specifically, in the first column the details (age, gender) of the anatomic human models used to extract the calf are given; the second and the third columns show the average B_1^+ magnitude calculated in the axial slice crossing the coil center and the maximum of the local SAR calculated in the same slice, when applying 1 W of RF power. In all the models the maximum of the local SAR occurs in axial slice crossing the coil center and in proximity of the calf surface and of the loop. The fourth column shows the maximum of the local SAR after scaling the simulations to achieve the same B_1^+ slice average value of 1μT.

Figure 2b, c shows the maps of B_1^+ magnitude [μT] and SAR [W/kg] in the axial slice crossing the eyes obtained by using CST MW Suite with the human head extracted from Ella. Both maps, which refer to continuous RF wave excitation, are given only in the region occupied by the load.

Table 4 summarizes the results obtained by using CST MW Suite for the human heads extracted from all the models listed in Table 1. Similarly to Table 3, in the first column the details (gender, age) of the anatomic human models used to extract the head are given; the second and the third columns show the average B_1^+magnitude calculated in the axial slice crossing the eyes and the maximum of the local SAR calculated in the entire head, when applying 1W of input power at each source. In this context, the maximum local SAR occurs away from the aforementioned axial slice: more in details, in Billie and Roberta the maximum local SAR occurs in the bottom of neck, in Ella and

Thelonious in the upper part of neck, in Hugo, Duke and Dizzy in the mandibular region. The fourth column shows the maximum local SAR after scaling the simulations to achieve the same B_1^+ slice average value of 1μT.

Figure 3b shows the measured B_1^+ magnitude [μT] (in the central axial plane) acquired on the scanner when the birdcage was loaded with the cylindrical phantom and using a total input power of 0.1151 kW: note that this value represents the total input power at the birdcage section, since it has been obtained after taking into account all the cable losses in the RF chain. Figure 3c shows the simulated B_1^+ magnitude [μT] obtained by using the same total input power of the measurement.

DISCUSSIONS

Excellent agreement can be observed between the measurement in Figure 3b and simulation in Figure 3c; the slight difference between peak values might be due to possible errors in the dielectric parameters used when simulating the 0.1 M saline solution. The asymmetry in Figure 3b is presumably due to the not-exact circular symmetry of the load (related to manufacturing reasons) and to the asymmetric effects inherent in feeding and matching [Tiberi et al., 2013].

B_1^+ map given in Figure 1 shows the typical quadpolar effect generated by a single loop [Robitaille et al., 2006]; moreover, B_1^+ decreased with depth into the calf. We noted that modification of the matching capacitor leads to a modification of the scale of both B_1^+ magnitude and SAR maps, but the shape of these maps is not affected. This is in agreement with what is reported in Kozlov et al. [2009], and it holds true if the effects of fringe fields of capacitors are negligible.

From the third column of Table 3, one might conclude that the maximum SAR is higher in smaller load sizes, i.e., in children; this is in agreement with Conil et al. [2008] where the SAR due to a plane wave has been investigated. From the second column, it is possible to note a large variability in the average B_1^+: this variability (quantified by a standard deviation of 0.38) depends on the different shape and size of the human calf models (i.e., different boundary conditions) and on the different loading (i.e., matching) conditions. Moreover, from the second column, it is possible to note that the average B_1^+ increases monotonically with decreasing load size; such behavior can be explained by recalling that the magnitude of B_1^+ decreases with depth into the calf and this is reflected in an average B_1^+ increase as the load size becomes smaller (being the distance between the calves and the center of the coil fixed in all the simulations). The same behavior is not reproduced in the third column where the maximum of the local SAR is given, since the SAR is related to the RF electric field and not to the RF magnetic field (note that here RF electric and magnetic fields are solutions to Maxwell equations in near-field region where plane wave assumptions do not hold). As explained in the previous section, the fourth column shows the maximum local SAR after scaling the simulations to achieve the same B_1^+ slice average value of 1μT: this is the type of scaling operation that is performed in MR to produce the desired B_1^+, irrespective of load. Obviously, simulations can be scaled to achieve an arbitrary B_1^+ slice average value (rather than 1μT) or an arbitrary average flip angle, similarly to what conventional routines for RF pulse calibration do in MR scanners to establish the B_1^+ field required to obtain the maximum MR signal. After such scaling, the maximum local SAR decreases with decreasing load size, i.e., in children.

Turning now to the second scenario, the B_1^+ map in Figure 2 shows the typical central focusing effect [Robitaille et al., 2006]. Again, we noted that modification of the matching capacitor leads to a modification of the scale of both B_1^+ magnitude and SAR maps, but the shape of these maps does not change.

From the third column of Table 4, one might conclude that the maximum SAR is higher in smaller load sizes, i.e., in children. From the second column, it is possible to note a variability in the average B_1^+, but such variability (quantified by a standard deviation of 0.06) is less pronounced with respect to the corresponding column of Table 3; this can be explained by noting that different head models lead to similar matching conditions. Moreover, differently from the previous scenario, the average B_1^+ does not increase monotonically with decreasing load size, and this is presumably due to the B_1^+ magnitude, which exhibits the central focusing effect for all human heads simulations. The fourth column of Table 4 shows the maximum local SAR after scaling the simulations to achieve the same B_1^+ slice average value of 1μT: from this column, it is not possible to confirm the trend of the corresponding column of Table 3. In more detail, the average of maximum local SAR in adults (i.e., Hugo, Duke, Ella) is 1.93 W/kg; the maximum of the local SAR in Thelonoius is 4% lower than the average of the maximum of the local SAR in adults, while the maximum of the local SAR in Billie is 15% higher than the average of the maximum of the local SAR in adults.

The variation of the dielectric properties of tissues with the age is not considered in this study; however, in Wang et al. [2006] it is pointed out that such variation does not affect significantly the SAR (i.e., less than 10% in the extreme case).

CONCLUSIONS

Maximum local SAR in UHF MR with respect to the load size has been investigated by scaling the simulations to achieve the same B_1^+ slice average value of 1μT, which is the type of scaling operation, i.e., power calibration, that is done in MR to obtain the desired B_1^+ irrespectively of load. It has been shown that, in the surface loop scenario, the maximum local SAR decreases with decreasing load size, i.e., in children; on the contrary, in the volume coil scenario, the maximum local SAR in children was up to 15% greater than that in adults. Based on these results, a safety factor for the maximum local SAR can be appropriately set when performing UHF MR in juvenile subjects.

Finally, we remind the reader that, to assess the compliance of a specific MR sequence with the legal requirements, a 6 min time average has to be considered, as specified in regulation number IEC 60601-2-33. More in detail, the legal requirements are: average SAR<3.2 W/kg; maximum local SAR in the extremities <20 W/kg; maximum local SAR in head <10 W/kg. Since the SAR depends on the characteristics of the pulse-sequence adopted during the MR acquisition, all the parameters related to the sequence itself have to be taken into account in the calculation, including the number, shape and length of RF pulses that generate the desired flip angles. Just to give an example, the fifth column of Table 4 shows the maximum local SAR calculated for a gradient recalled echo (GRE) sequence (a sequence commonly used for 7.0 T head imaging) having the following parameters: flip angle= 90°, RF pulse length =3.2 ms, RF pulse shape: truncated-sinc; matrix size=512×512; sequence length=256 s; number of slices=12; frequency field of view=1, number of acquisitions=1.

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Figure Captions

Figure. 1. a) Single loop (external radius of 5 cm) near the HUGO calf; the axis of the loop is perpendicular to z-axis. The cones displayed along the loop indicate the source (bottom cone) and the tuning and matching capacitors. B_1^+ (b) and SAR (c) maps (when applying 1W of input power) in the z-plane, i.e., axial slice, crossing the loop center; the loop, shown in dotted black line, is 15 mm from calf surface. B_1^+ is in μT, SAR in W/Kg. In the HUGO calf simulation, approximately 4 million mesh nodes have been used (simulation time= 16 h on 1 workstation). The quadpolar effect (typical for the single loop) is clearly visible in B_1^+ map.

Figure. 2. a) Ella head inside the MR quadrature birdcage coil (accessible radius of 29.5 cm). The 4 cones displayed in the superior part indicate the 4 sources. The gaps in the two end-rings denote the location of tuning and matching capacitors (not displayed); B_1^+ (b) and SAR (c) maps (when applying 1W of input power at each source) in the z-plane, i.e., axial slice, crossing the eye of Ella center. B_1^+ is in μT, SAR in W/kg. In Ella head simulation, approximately 12.5 million mesh nodes have been used (simulation time= 15 h on 1 workstation using the GPU computing).The central focusing effect (typical for the birdcage) is clearly visible in B_1^+ map.

Figure. 3. a) Quadrature birdcage coil, described in the previous figure, loaded with 0.1 M saline solution cylindrical phantom (radius of 6 cm, height of 24 cm); b) B_1^+ magnitude in the central z-plane, i.e., central axial slice, measured by using the BSS method; c) B_1^+ magnitude obtained through simulation. The central focusing effect (typical for the birdcage) is clearly visible in B_1^+ maps. B_1^+ is in μT