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► To cite this version:

Julien Bordes, Sébastien Incerti, Erick Mora-ramirez, Jonathan Tranel, Cédric Rossi, et al.. Monte Carlo dosimetry of a realistic multicellular model of follicular lymphoma in a context of radioimmunotherapy. *Med.Phys.*, 2020, mp.14370. 10.1002/mp.14370 . hal-02981284

HAL Id: hal-02981284

<https://hal.science/hal-02981284v1>

Submitted on 8 Nov 2020

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1 **Monte Carlo dosimetry of a realistic multicellular model of follicular lymphoma in a**
2 **context of radioimmunotherapy**

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30 *Keywords:*

31 Small-scale dosimetry, realistic biological data, rituximab, Auger electrons, β^- particles.

32 *Abstract:*

33 *Purpose.* Tumors geometry and radiopharmaceutical biodistribution impact the energy deposi-
34 tion in targeted radionuclide therapy. However, small-scale dosimetry studies generally consider an ar-
35 tificial environment where the tumors are spherical and the radionuclides are homogeneously biodis-
36 tributed. To bring realism, we developed a dosimetric methodology based on a three-dimensional *in*
37 *vitro* model of follicular lymphoma incubated with rituximab, an anti-CD20 monoclonal antibody used
38 in the treatment of non-Hodgkin lymphomas, which might be combined with a radionuclide. The effects
39 of the realistic geometry and biodistribution on the absorbed dose were highlighted by comparison with
40 literature data. Additionally, to illustrate the possibilities of this methodology, the effect of different ra-
41 dionuclides on the absorbed dose distribution delivered to the *in vitro* tumor were compared.

42 *Methods.* The starting point was a model named Multicellular Aggregates of Lymphoma Cells
43 (MALC). Three MALCs of different dimensions and their rituximab biodistribution were considered. Ge-
44 ometry, antibody location and concentration were extracted from selective plane illumination micros-
45 copy. Assuming antibody radiolabeling with Auger electron (^{125}I and ^{111}In) and β^- particle emitters
46 (^{177}Lu , ^{131}I and ^{90}Y), we simulated energy deposition in MALCs using two Monte Carlo codes: Geant4-
47 DNA with “CPA100” physics models for Auger electrons emitters and Geant4 with “Livermore” physics
48 models for β^- particles emitters.

49 *Results.* MALCs had ellipsoid-like shapes with major radii, r , of ~ 0.25 , ~ 0.5 and ~ 1.3 mm. Ritux-
50 imab was concentrated in the periphery of the MALCs. The absorbed doses delivered by ^{177}Lu , ^{131}I and
51 ^{90}Y in MALCs were compared with literature data for spheres with two types of homogeneous biodistri-
52 butions (on the surface or throughout the volume). Compared to the MALCs, the mean absorbed doses
53 delivered in spheres with surface biodistributions were between 18% and 38% lower, while with vol-
54 ume biodistribution they were between 15% and 29% higher. Regarding the radionuclides comparison,
55 the relationship between MALC dimensions, rituximab biodistribution and energy released per decay
56 impacted the absorbed doses. Despite releasing less energy, ^{125}I delivered a greater absorbed dose per
57 decay than ^{111}In in the $r \sim 0.25$ mm MALC ($6.78 \cdot 10^{-2}$ vs. $6.26 \cdot 10^{-2} \mu\text{Gy} \cdot \text{Bq}^{-1} \cdot \text{s}^{-1}$). Similarly, the absorbed
58 doses per decay in the $r \sim 0.5$ mm MALC for ^{177}Lu ($2.41 \cdot 10^{-2} \mu\text{Gy} \cdot \text{Bq}^{-1} \cdot \text{s}^{-1}$) and ^{131}I ($2.46 \cdot 10^{-2} \mu\text{Gy} \cdot$
59 $\text{Bq}^{-1} \cdot \text{s}^{-1}$) are higher than for ^{90}Y ($1.98 \cdot 10^{-2} \mu\text{Gy} \cdot \text{Bq}^{-1} \cdot \text{s}^{-1}$). Furthermore, radionuclides releasing more
60 energy per decay delivered absorbed dose more uniformly thorough the MALCs. Finally, when consid-
61 ering the radiopharmaceutical effective half-life, due to the biological half-life of rituximab being best
62 matched by the physical half-life of ^{131}I and ^{177}Lu compared to ^{90}Y , the first two radionuclides delivered
63 higher absorbed dose.

64 *Conclusion.* In the MALCs considered, β^- emitters delivered higher and more uniform absorbed
65 dose than Auger electron emitters. When considering radiopharmaceutical half-lives, ^{131}I and ^{177}Lu de-
66 livered absorbed doses higher than ^{90}Y . In view of real irradiation of MALCs, such a work may be useful
67 to select suited radionuclides and to help explain the biological effects.

68 1 Introduction

69 In targeted radionuclide therapy (TRT), a tumor-seeking agent is combined with an emitter of
70 short-range radiation, such as Auger electrons (typically inferior to 1 μm in water), α particles (inferior
71 to 100 μm) and β - particles (in the order of millimeters). The objective of this combination is to deliver
72 in patients a lethal absorbed dose to tumor cells while minimizing the exposure of healthy tissues. Thus,
73 the development and the evaluation of TRT requires small-scale dosimetry studies^{1,2}. However, previous
74 studies have been generally based on simplifying assumptions (*e.g.* ³⁻⁸) owing to the lack of data at the
75 submillimeter scale regarding the tumor geometry and the radionuclide biodistribution.

76 Firstly, tumors were generally supposed spherical, or, alternatively, ellipsoidal. Absorbed dose
77 in spheres was found to be higher than in the ellipsoids of the same volume⁹, emphasizing the influence
78 of the geometry. Secondly, small-scale dosimetry calculations have regularly assumed a homogeneous
79 radionuclide biodistribution, either at the surface or within the volume of the tumor. In practice, the
80 targeting of a tumor is not homogeneous, partly due to the heterogeneous distribution of the receptors
81 (*i.e.* the antigens). Thus, mathematical models of heterogeneous radionuclide biodistributions within
82 the tumors have been studied¹⁰⁻¹³. As could be expected, these models led to absorbed doses signifi-
83 cantly less uniform within the tumor compared to homogeneous biodistributions. Therefore, given the
84 impact of both tumor geometry and radionuclide biodistribution, there are clear calls for the develop-
85 ment of small-scale dosimetry based on realistic biological data.

86 In the context of radioimmunotherapy (RIT), a single *in vitro* cultured carcinoma cell was con-
87 sidered in order to reproduce the effect of its realistic geometry and monoclonal antibody biodistribu-
88 tion on the dosimetry¹⁴. These biological data were extracted from microscope images. Assuming anti-
89 body radiolabeling with iodine-125, the energy deposition in the cell was calculated with Monte Carlo
90 simulations. However, the relevance of this work was limited since tumors targeted in RIT such as non-
91 Hodgkin lymphoma (NHL) grow as multicellular three-dimensional structures. Thus, a single cell dosi-
92 metric model neglects the absorbed dose delivered by the radiopharmaceutical bound to other cells of
93 the tumor (crossfire effect). Furthermore, studies regarding unlabeled monoclonal antibody treatments
94 highlighted that the biodistribution might be impacted by the tumor size and the extracellular matrix

95 (*i.e.* the assemblage of non-cellular molecules providing structural support for the surrounding cells),
96 limiting the effectiveness of the treatment. Increasing tumor size is regarded as possibly reducing the
97 monoclonal antibody concentration and decreasing survival of a mouse model¹⁵ and patients¹⁶. Further-
98 more, the extracellular matrix might hamper the monoclonal antibody penetration¹⁷. To address these
99 questions, a three-dimensional *in vitro* culture system of follicular lymphoma (a NHL subtype) express-
100 ing extracellular matrix was designed in a separate project at the Cancer Research Center of Toulouse¹⁸⁻
101 ²¹. The model was termed multicellular aggregate of lymphoma cells (MALC). The MALCs were incu-
102 bated with rituximab. This chimeric monoclonal antibody targeting the membrane-associated CD20 an-
103 tigen has been combined with radionuclides for RIT treatments (see clinical review of Read *et al.*²²).

104 To improve the realism and the relevance of the single cell dosimetry, this paper presents a do-
105 simetric model based on selective plane illumination microscopy (SPIM) images of three MALCs, of dif-
106 ferent size, treated by rituximab. Data were extracted to define the tumor geometry and the radiophar-
107 maceutical biodistribution. Rituximab was virtually radiolabeled with two types of electron emitters.
108 The Auger electron emitters iodine-125 (¹²⁵I) and indium-111 (¹¹¹In) were studied because they proved
109 their efficiency in destroying lymphoma cells *in vitro* in RIT-related research^{23,24}. β^- particle emitters
110 lutetium-177 (¹⁷⁷Lu), iodine-131 (¹³¹I) and yttrium-90 (⁹⁰Y) were also considered. For the RIT treatment
111 of patients with relapsed / refractory NHL, ¹⁷⁷Lu-lilotomab satetraxetan is currently on trial²⁵, while
112 both ¹³¹I-tositumomab (Bexxar) and ⁹⁰Y-ibritumomab tiuxetan (Zevalin) have been used²⁶. In RIT, the
113 dosimetry is highly impacted by the radiopharmaceutical effective half-life, a parameter that would de-
114 pend on the rituximab radiolabeling. We studied the energy deposition patterns in MALCs with Monte
115 Carlo modeling of radiation transport. Additionally, the effective half-life was accounted for by using the
116 simulated data and a simple pharmacokinetic model.

117 As a first step, the relevance of implementing realistic biological data was highlighted by com-
118 paring average absorbed doses to literature data. For the following step, to illustrate the possibilities
119 offered by this new dosimetric model, we compared the radionuclides' absorbed fraction, average ab-
120 sorbed dose and absorbed dose volume distribution. As a last step, the influence of the radiopharma-
121 ceuticals half-life on the absorbed dose was studied.

122 2 Material and methods

123 2.1 Three-dimensional model of follicular lymphoma

124 The development of the MALCs as well as the microscopy acquisition were detailed in prior stud-
125 ies¹⁸⁻²¹. This section focuses on information relevant to the present work.

126 2.1.1 MALC growth and labeling

127 To compare the antibody biodistribution in different tumor volumes, the MALCs, created with
128 follicular lymphoma cells expressing green fluorescent protein (GFP), were cultured during 2, 5 or 10
129 days (hereafter referred to as D2, D5 and D10). Then, they were treated by a saturating concentration
130 of 10 µg/ml of CF555 labeled rituximab²⁷ for a duration of 24 h.

131 2.1.2 SPIM imaging

132 Fluorescence microscopy technologies, such as SPIM detect the light emitted by fluorescent
133 markers²⁸. SPIM was particularly adapted to this study since it allows the observation of whole organ-
134 isms or three-dimensional multicellular models over a few millimeters while keeping a low phototoxi-
135 city^{29,30}. 24 h after rituximab administration, z-stack of images were acquired. Each image corresponded
136 to a slice of the MALC in the x-y plane. For example, Figure 1 displays a central slice of MALCs (a) D2, (b)
137 D5 and (c) D10. Green signal corresponds to GFP 488 nm emission and reveals MALC geometry. The
138 black hollow at the center of MALCs D5 and D10 are due to a lack of follicular lymphoma cells. Red signal
139 from the 561 nm emission of CF555 represents the rituximab biodistribution. To optimize the contrast
140 between follicular lymphoma cells and rituximab, the colors on the images differ from the real colors
141 produced by the fluorescent markers. SPIM images were composed of voxels with (x, y and z) dimen-
142 sions of $3.7 \cdot 3.7 \cdot 3.1 \mu\text{m}^3$ and were stored into TIFF files.

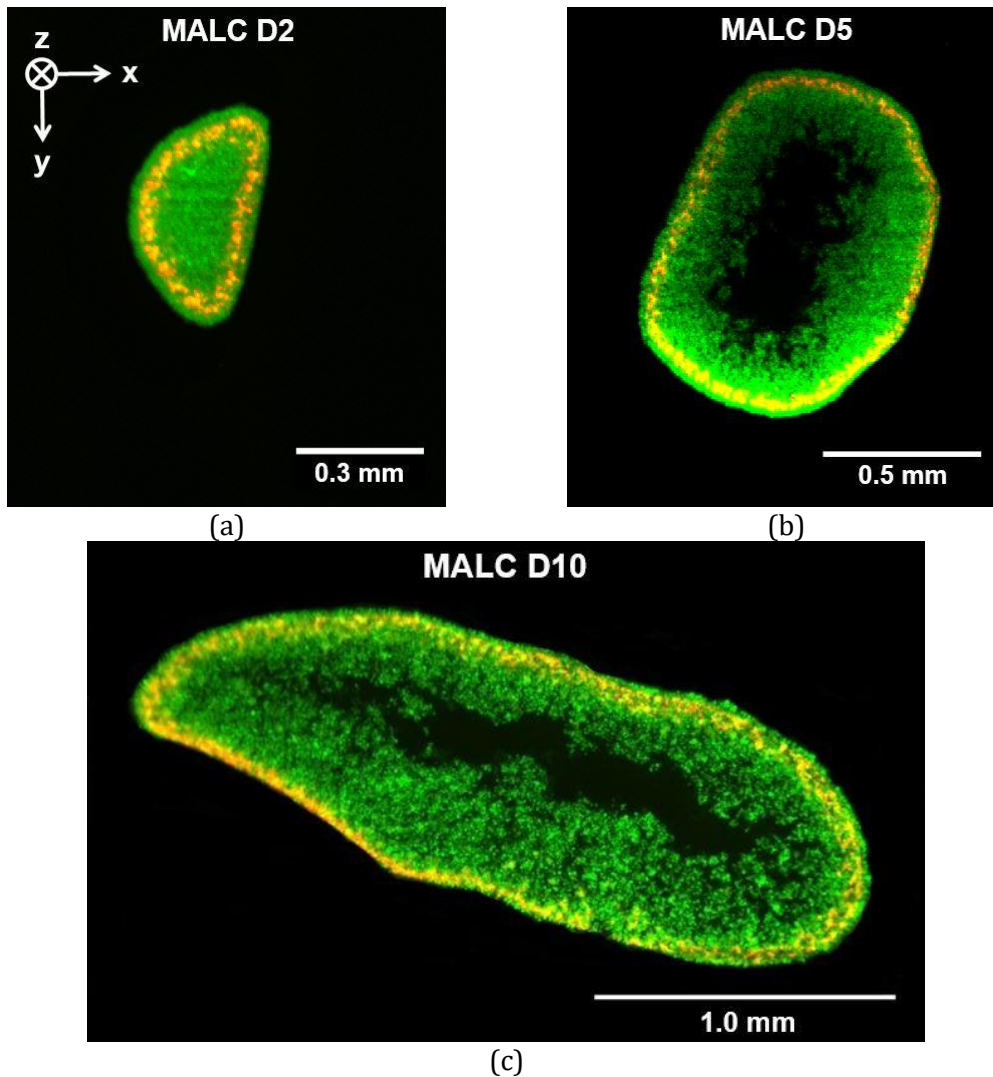


Figure 1. SPIM images of the central slices of the MALC: (a) D2, (b) D5 and (c) D10. Follicular lymphoma cell nuclei and rituximab appear in green and red, respectively.

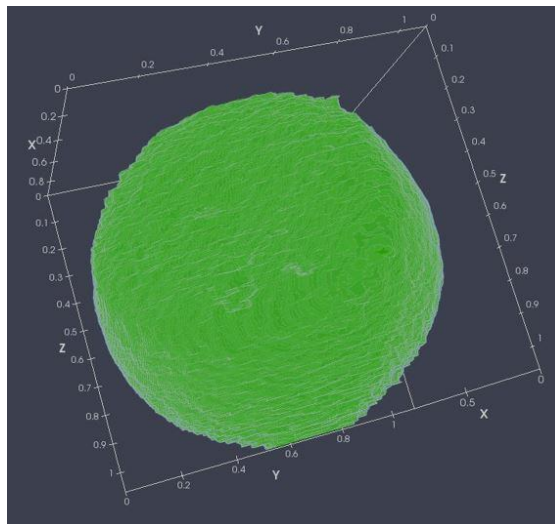
143 2.2 Image processing

144 To extract the data required for the dosimetry of the MALCs, SPIM images were processed with
 145 Fiji³¹ (“Fiji Is Just ImageJ”, downloadable from fiji.sc). This program has specific features for the
 146 processing of microscope images of three-dimensional biological samples³² and can read TIFF files. First,
 147 the stacks of images were transferred to Fiji for 3D reconstructions (*e.g.* MALC D5 in Figure 2). Then,
 148 two macros were developed and implemented in Fiji:

149 - The first one treated the green signal in order to extract the MALCs geometry. MALC voxels
 150 were separated from the background by applying an intensity threshold. A clear delineation of the
 151 MALCs boundary allowed a straightforward determination of the threshold value. Since the black hollow
 152 sections of the MALC D5 and D10 were in the center, they were considered as a part of the volume of
 153 interest (VOI). Finally, in view of the definition of the MALC geometry during Monte Carlo simulations,

154 the macro recorded different information into a “.dat” file; namely, the number of voxels, the voxel di-
155 mensions along the three axes, and each voxel coordinate (x, y, z).

156 - The second macro treated the red signal from the rituximab CF555 label. It removed the back-
157 ground noise, then recorded the voxel coordinates and intensity into a “.dat” file. During simulations,
158 the source voxels were selected with a probability directly derived from their red signal intensity. It was
159 supposed that the CF555 biodistribution would be identical to the radiopharmaceutical. This assumes
160 that the CF555 labeling would neither detach from the antibody nor impact its biodistribution. To the
161 best of our knowledge, no publication addresses these questions. This also assumes that the radiolabel-
162 ing would not affect the rituximab biodistribution. This last point is supported by the requirement that,
163 in nuclear medicine, the radiolabeling strategies preserve the biological properties of the proteins³³ such
164 as rituximab.



165

166 Figure 2. 3D reconstruction of MALC D5 from SPIM images. The axes are given in millimeters.

167

167 2.3 Voxel-based dosimetry

168 2.3.1 General settings

169 Energy depositions were calculated with the Geant4 Monte Carlo open-source platform³⁴⁻³⁶
170 since it enables the implementation of voxelized geometries and its physics models are well adapted to
171 energies encountered in TRT. Furthermore, Geant4, being a well-established object-oriented code base,
172 has a high degree of reuse meaning that little code must be written for a new application. The Geant4

173 “microbeam” example served as a basis. It was developed at the Centre d’Études Nucléaires de Bor-
174 deaux-Gradignan (CENBG) to simulate the irradiation of individual cells with a microbeam of protons
175 or α particles^{37,38}. In order to suit this example for TRT, we added the possibility of simulating the irra-
176 diation of a VOI (in this work the MALC) by internalized radiopharmaceuticals. To reconstruct the
177 MALCs, the class *G4PVParameterised* placed the SPIM images voxels ($3.7 \cdot 3.7 \cdot 3.1 \mu\text{m}^3$) according to
178 their coordinates stored in the output file of the first macro (see 2.2). The MALCs were positioned into
179 a sphere of liquid water. This enabled electron backscattering into the MALCs with the consequence of
180 an additional energy deposition. The simulation of the irradiation of each MALC by each radionuclide
181 considered 10^6 decays. The activity was assumed homogeneously biodistributed within a voxel source.
182 Accordingly, the coordinates of a decay site inside such a voxel were sampled from a uniform distribu-
183 tion. The particles were then emitted isotropically. Depending on the type of electron emitter, energy
184 depositions in the MALCs were simulated with different methods that will be presented in the following
185 sections (2.3.2 and 2.3.3).

186 **2.3.2 Auger electron emitters**

187 ^{125}I and ^{111}In mean emission spectra of monoenergetic electrons (Auger and internal conversion
188 electrons, Figure 3) and photons (X-rays and γ -rays) were extracted from the publication of the Ameri-
189 can Association of Physicists in Medicine³⁹ (AAPM), as it is still a reference for Auger electron emitters
190 (*e.g.* ^{14,40,41}). Table 1 shows important properties of the decay of ^{125}I and ^{111}In . Half-lives were extracted
191 from the Medical Internal Radiation Dose (MIRD) book⁴². The mean penetration corresponds to the
192 length of the vector connecting the electron emission and stopping points. Hence, this distance is lower
193 than the path length because the latter considers the direction changes of the electrons. Although nu-
194 clear medicine dosimetric studies usually report the path length, in the present work the penetration
195 was more relevant for analyzing the results.

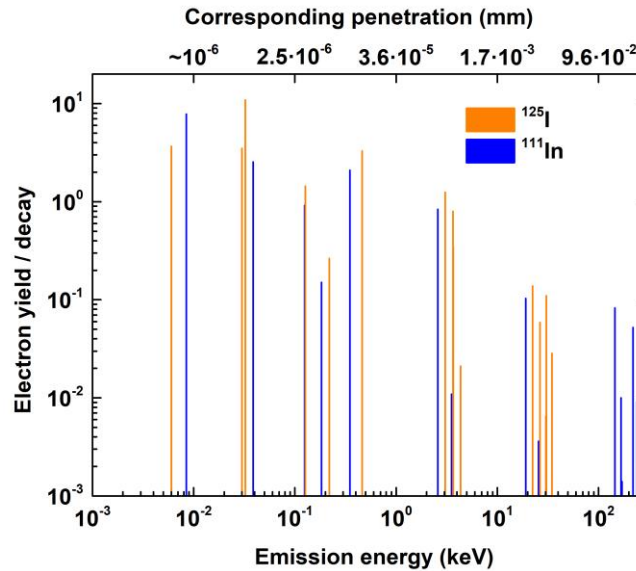


Figure 3. Mean electron emission spectra (Auger and internal conversion electrons) of ^{125}I and ^{111}In from the AAPM publication³⁹. For each energy decade, the top x-axis provides the corresponding mean penetration according to Geant4-DNA.

Table 1. Some physical properties of the Auger electron emitters. Half-lives come from MIRD data⁴², spectra data were extracted from the AAPM publication³⁹ and mean penetrations were determined with Geant4-DNA.

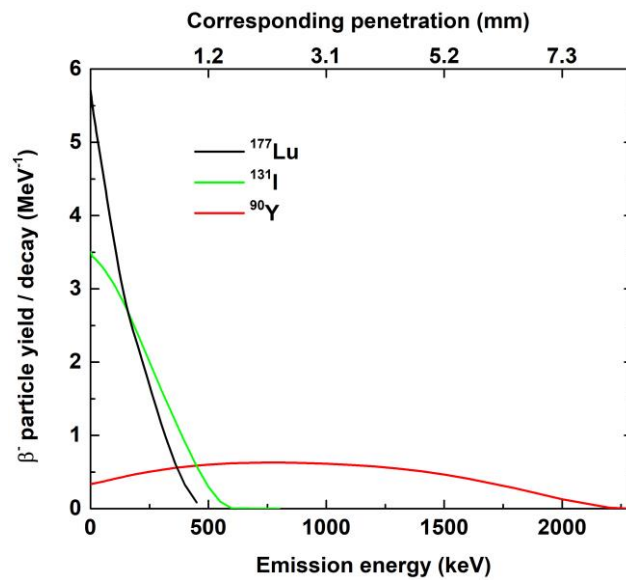
Auger electron emitter	^{125}I	^{111}In
Half-life (day)	59.4	2.8
Mean electron energy released (keV/decay)	19.5	34.7
Electron yield per decay	25.8	14.9
Electron mean energy $E_{e^- \text{ mean}}$ (keV)	0.8	2.3
Penetration at $E_{e^- \text{ mean}}$ (mm)	$2.5 \cdot 10^{-5}$	$1.4 \cdot 10^{-4}$
Maximum energy $E_{e^- \text{ max}}$ (keV)	34.7	245.0
Penetration at $E_{e^- \text{ max}}$ (mm)	$1.5 \cdot 10^{-2}$	$4.2 \cdot 10^{-1}$

Auger electrons induce energy deposition clusters localized at the nanometer scale⁴³. Such phenomena can be studied with step-by-step Monte Carlo codes since they give the track structure of the particles down to the nanometer scale. Accordingly, ^{125}I and ^{111}In emissions were simulated with Geant4-DNA⁴⁴⁻⁴⁷ (geant4-dna.org). “CPA100” physics models (“option 6”)^{48,49} were used. They describe ionization, electronic excitation and elastic scattering processes and are applicable from 256 keV down to 11 eV in liquid water.

210 For the simulations of photon interactions (Compton and Rayleigh scatterings as well as photo-
 211 electric absorption), Geant4-DNA does not provide any physics models to-date. Thus, Geant4 “Liver-
 212 more” physics models for photons were selected^{50,51}. During simulations, electrons and photons could
 213 produce vacancies in the atomic shells so atomic deexcitation emission was tracked (Auger electron
 214 emission, Auger cascade and fluorescence).

215 2.3.3 β^- particle emitters

216 AAPM’s publication³⁹ only provides data for Auger electron emitters. Therefore, ^{177}Lu , ^{131}I and
 217 ^{90}Y mean emission spectra of β^- particles (Figure 4), monoenergetic electrons and photons were ex-
 218 tracted from the MIRDB book⁴². Table 2 presents the main properties of the decay of ^{177}Lu , ^{131}I and ^{90}Y .



219

220 Figure 4. β^- particle emission spectra of ^{177}Lu , ^{131}I and ^{90}Y from MIRDB book⁴². The top x-axis provides the corre-
 221 sponding Geant4 mean penetration every 500 keV.

222

223 Table 2. Some physical properties of the β^- particle emitters. Half-lives and spectra data come from MIRD book⁴²
 224 and mean penetrations were determined with Geant4.

β^- particle emitter	¹⁷⁷ Lu	¹³¹ I	⁹⁰ Y
Half-life (day)	6.6	8.0	2.7
Mean electron energy (β^- particles + monoenergetic electrons) released (keV/decay)	147.8	191.7	933.2
β^- particle mean energy $E_{\beta^- \text{ mean}}$ (keV)	133.1	181.8	933.0
Penetration at $E_{\beta^- \text{ mean}}$ (mm)	$1.6 \cdot 10^{-1}$	$2.6 \cdot 10^{-1}$	2.9
β^- particle maximum energy $E_{\beta^- \text{ max}}$ (keV)	497.8	806.9	2280.1
Penetration at $E_{\beta^- \text{ max}}$ (mm)	1.2	2.4	8.4

225

226 Part of the emission of ¹⁷⁷Lu, ¹³¹I and ⁹⁰Y has an energy too high to be modeled with Geant4-DNA
 227 physics models. Thus, simulations of the irradiation with these emitters were accomplished with Geant4
 228 in a condensed history mode, as this provides physics models applicable up to GeV energies. “Liver-
 229 more”^{52,53} physics models were used to describe particles interactions.

230 The ionization process was simulated along with bremsstrahlung. This latter slightly increased
 231 the stopping power of the electrons / β^- particles at these energies, thus marginally reducing their pen-
 232 etration. “Livermore” electron models are defined down to about 10 eV. Nevertheless, in accordance
 233 with guidelines, the cut-off was set to 250 eV. Energy production cuts for secondary particles was also
 234 fixed at 250 eV. In order to limit the number of electrons crossing a voxel with no energy deposition,
 235 step size limit was arbitrary chosen to 0.6 μm , corresponding to less than 1/5 of the voxel dimensions.

236 2.4 Dosimetric quantities

237 2.4.1 Simulation results

238 Geant4 simulations generated output files containing the energy deposited in the MALCs voxels.
 239 Firstly, different dosimetric quantities were directly calculated from these data: the absorbed fraction
 240 and the absorbed dose.

241 - The absorbed fraction ϕ represents the fraction of energy released from the source region that
 242 is deposited in the target region. Here, the MALCs represent both the source and the target regions. The
 243 results focused on electron absorbed fractions ϕ_{e^-} (monoenergetic electrons and eventually β^- parti-
 244 cles), because photon absorbed fractions are insignificant at this scale:

$$\phi_{e^- \text{ MALC}} = \frac{E_{e^-}}{E_{e^-_0}} \quad (1)$$

245 Where E_{e^-} is the deposited energy and $E_{e^-_0}$ the electron-released energy. The absorbed fraction
 246 is frequently reported in nuclear medicine dosimetry studies⁵⁴.

247 - The energy deposited E_i (in J) in a VOI i (either a whole MALC or a voxel) characterized by a
 248 mass m_i (in kg) was converted to absorbed dose D_i (in Gy):

$$D_i = \frac{E_i}{m_i} \quad (2)$$

249 This quantity was then divided by 10^6 to obtain the mean absorbed dose per decay (or per unit
 250 of time-integrated activity, in $\text{Bq} \cdot \text{s}$, according to MIRD most recent terminology⁵⁵). First, this quantity
 251 was determined for the whole MALCs. However, the absorbed dose was not uniformly delivered within
 252 the volume. In order to specify the volume fractions having received either low or high absorbed dose,
 253 cumulative dose-volume histograms (cDVH) were plotted. They give the fraction of the volume (*i.e.* the
 254 fraction of MALC voxels) in which an absorbed dose greater than or equal to a given value was delivered.
 255 These graphs are routinely used in external radiotherapy, but also in some TRT studies^{56,57}.

256 2.4.2 Considering radiopharmaceutical effective half-life

257 Subsequently, in order to consider the effective half-life of the radiopharmaceutical X-rituximab
 258 (X being one of the radionuclides), the MALCs and voxels absorbed doses per unit administered activity
 259 (A_0 in Bq) were calculated. In nuclear medicine, this quantity allows the assessment of the absorbed
 260 dose delivered to a VOI for a given administered activity. It was calculated as the product of the absorbed
 261 doses per decay (see 2.4.1) and the time-integrated activity coefficient \tilde{a} (also known as the residence
 262 time and given in s). The value of \tilde{a} represents the total number of decays occurring in a MALC per unit
 263 administered activity⁵⁵:

$$\tilde{a} = \frac{1}{A_0} \int_0^{\infty} A(X - \text{rituximab}, t) \cdot dt \quad (3)$$

264 Where $A(X - \text{rituximab}, t)$ is the time-dependent activity of the radiopharmaceutical in a MALC
 265 (in Bq). Assuming that this radiopharmaceutical has an instantaneous uptake in a MALC (*i.e.* instanta-
 266 neously bound to the MALCs according to the biodistribution extracted from the images) and its elimi-
 267 nation is monoexponential:

$$\tilde{a} = \frac{1}{A_0} \int_0^{\infty} A_0 \cdot e^{-(\ln(2)/T_{\text{eff}}(X - \text{rituximab})) \cdot t} \cdot dt \quad (4)$$

$$\tilde{a} = 1.443 \cdot T_{\text{eff}}(X - \text{rituximab}) \quad (5)$$

268

269 Where $T_{\text{eff}}(X - \text{rituximab})$, the effective half-life of the radiopharmaceutical (in s), depends on $T_{\text{phys}}(X)$
 270 the physical half-life of the radionuclide X (in s) and on $T_{\text{bio}}(\text{rituximab})$ the biological half-life of rituxi-
 271 mab (in s):

$$T_{\text{eff}}(X - \text{rituximab}) = \frac{T_{\text{phys}}(X) \cdot T_{\text{bio}}(\text{rituximab})}{T_{\text{phys}}(X) + T_{\text{bio}}(\text{rituximab})} \quad (6)$$

272

273 The physical half-lives are well documented (see Table 1 and Table 2), but, rituximab biological
 274 half-life in NHL is less-well established. In a clinical study, Scheidhauer *et al.* evaluated the effective half-
 275 life of the radiopharmaceutical ^{131}I -rituximab in NHL⁵⁸. Biodistribution in 14 patients was determined
 276 by gamma camera imaging. Our work focused on the 3.4 days mean effective half-life determined by
 277 Scheidhauer *et al.* We emphasize, however, that its precision is limited. It is associated to a standard
 278 deviation of 1.1 day. Also, Scheidhauer *et al.* reported a high inter-patient variability, with effective half-
 279 lives ranging from 1.7 to 5.5 days. The gamma camera imaging introduces further uncertainties on the
 280 dosimetry⁵⁹⁻⁶⁰. According to equation (6), the 3.4 days effective half-life corresponds to a rituximab bio-
 281 logical half-life of 5.8 days. Using this value and the equation (6), the effective half-lives of rituximab
 282 radiolabeled with ^{125}I , ^{111}In , ^{177}Lu or ^{90}Y were deducted.

283 3 Results

284 3.1 MALC and rituximab biodistribution characteristics

285 Table 3 presents the dimensions of the three MALCs along x-, y-, z- axes and their masses in
 286 Geant4 simulations which were used to determine the absorbed doses according to equation (2). The
 287 masses were evaluated by multiplying the number of voxels in a MALC by the mass of a voxel filled with
 288 liquid water (as simulations were performed in this material). Figure 1 shows that MALCs considered
 289 in this study have ellipsoid-like shape. For MALC D2, D5 and D10, major radii are ~ 0.25 mm, ~ 0.5 mm
 290 and ~ 1.3 mm, respectively. Rituximab is heterogeneously biodistributed inside a shell-like volume at
 291 the periphery of the MALCs (Figure 1). Generally, the shell-like volumes are characterized by a thickness
 292 and a distance to the surface of the MALCs both of the order of a few hundredths of a millimeter. It was
 293 previously demonstrated that the extracellular matrix limits the diffusion of therapeutic agents in tu-
 294 mors¹⁷. It was also shown that MALCs expressed extracellular matrix proteins²⁰. Accordingly, in an on-
 295 going MALC-based study, the possibility that the rituximab peripheral location is due to the extracellular
 296 matrix is being investigated.

297 Table 3. Sizes along x-, y- and z-axes of the MALCs D2, D5 and D10.

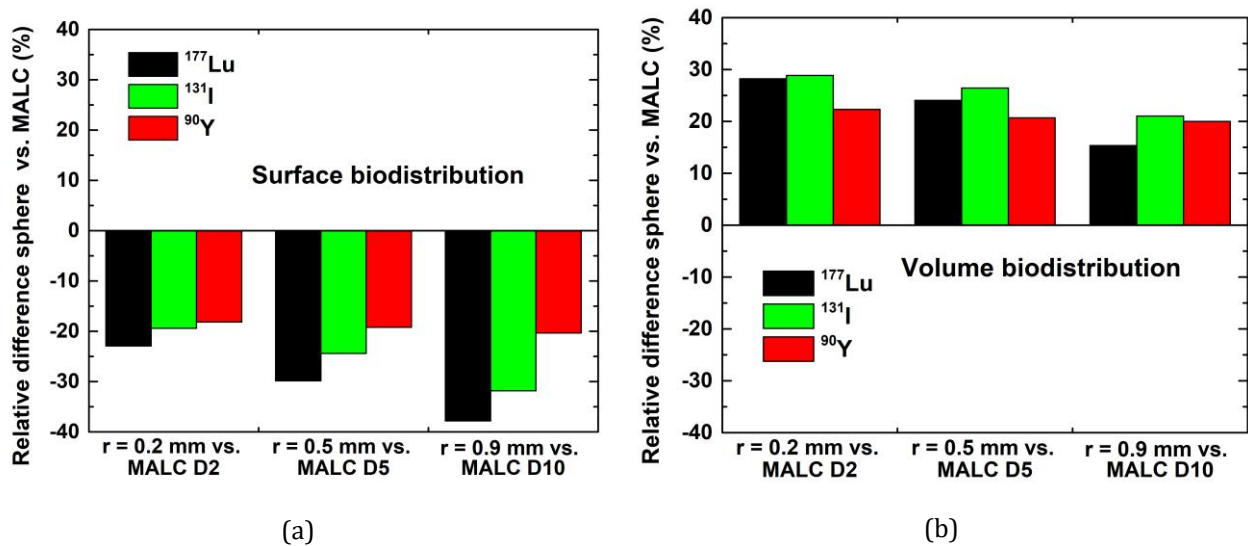
MALC	Dimensions x · y · z (mm ³)	Mass (mg)
D2	0.34 · 0.53 · 0.50	$4.6 \cdot 10^{-2}$
D5	0.96 · 1.09 · 1.07	$5.5 \cdot 10^{-1}$
D10	2.54 · 1.23 · 2.26	2.8

298 3.2 Dosimetric quantities per decay

299 3.2.1 Comparison with literature data obtained in a simplified model

300 Figure 5 displays the relative difference (%) of absorbed dose per decay between the reference
 301 data of Bardiès and Chatal³ for spheres with either (a) a surface or (b) a volume biodistribution and
 302 MALCs calculations (relative difference = $100 \cdot (D_{\text{sphere}} - D_{\text{MALC}}) / D_{\text{MALC}}$). In this previous study, calcula-
 303 tions were carried out by an analytical method based on scaled dose-point kernels in spheres with ra-
 304 dius ranging from 0.01 to 22 mm. For a valid comparison, the results for a MALC were compared to the
 305 results of the sphere having the closest mass. Accordingly, absorbed doses for MALCs D2, D5 and D10

306 were compared to the values for spheres with radius, r , of 0.2 mm, 0.5 mm and 0.9 mm, respectively.
 307 Furthermore, only ^{177}Lu , ^{131}I and ^{90}Y were considered because Bardiès and Chatal focused their study
 308 on β^- particle emitters. The peripheral biodistribution of rituximab (discussed in section 3.1) can be seen
 309 as an intermediate between the surface and volume biodistributions assumed by Bardiès and Chatal.
 310 Hence, this is coherent that the absorbed doses per decay for MALCs lie between the values for these
 311 two theoretical situations. It is also noticeable that results with these two biodistributions own a similar
 312 degree of agreement with results for the MALCs. This indicates that for the purpose of absorbed dose
 313 calculations, in these particular cases, the surface and the volume biodistribution are both equally rele-
 314 vant. Furthermore, non-negligible discrepancies between the theoretical model and realistic data were
 315 observed. These results highlight the need to account of realistic biological data.



316 Figure 5. Relative differences (%) between absorbed dose per decay calculated in spheres of various radii with
 317 either a (a) surface or (b) a volume biodistribution³ and MALCs (taken as reference). Comparison spheres vs.
 318 MALCs are as follow: $r = 0.2 \text{ mm vs. MALC D2}$ (left), $r = 0.5 \text{ mm vs. MALC D5}$ (middle) and $r = 0.9 \text{ mm vs. MALC}$
 319 D10 (right).

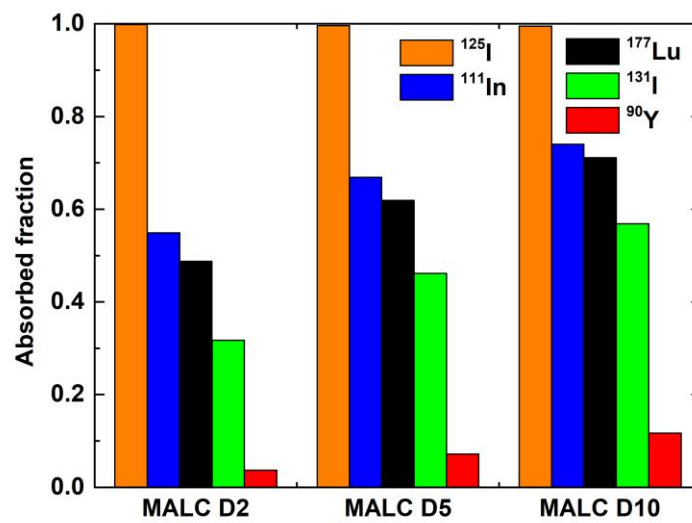
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321 3.2.2 Comparison of the radionuclides

322 For clarity, the results for each radionuclide are reported by ascending order of energy released
 323 per decay, that is: ^{125}I , ^{111}In , ^{177}Lu , ^{131}I and ^{90}Y .

324 Figure 6 displays, for each radionuclide, the electron absorbed fraction in the MALCs and Table
 325 4 shows the corresponding numerical values. As MALCs get bigger, the electrons / β^- particles deposit

326 more energy inside before exiting them, increasing the absorbed fractions. Assuming a homogeneous
 327 surface biodistribution of rituximab (which is at first sight the closer approximation of the real biodis-
 328 tribution as seen on Figure 1), for very large MALCs the absorbed fraction would tend toward 0.5 (semi-
 329 infinite medium condition). However, for MALC D5 and D10, the values become greater than 0.5 for all
 330 radionuclides but ^{90}Y (presumably, absorbed fraction of ^{90}Y would become greater than 0.5 in bigger
 331 MALCs). This can be explained by the rituximab penetration inside the MALCs, enabling the electrons /
 332 β^- particles to systematically deposit energy in the VOI, regardless their trajectories. This point is critical
 333 since it further demonstrates the importance of using realistic biological data.



334 Figure 6. Absorbed fractions in MALCs D2, D5 and D10 for ^{125}I , ^{111}In , ^{177}Lu , ^{131}I and ^{90}Y .
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337 Table 4. Absorbed fraction in the MALCs for the five radionuclides.

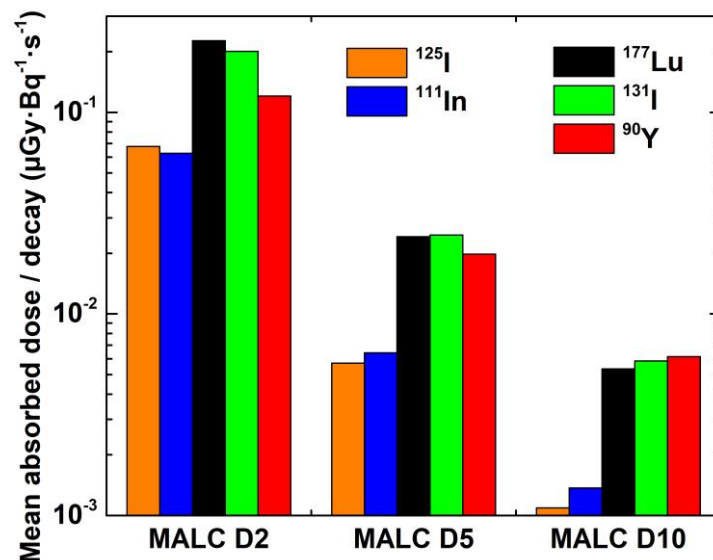
Radionuclide	Absorbed fraction		
	MALC D2	MALC D5	MALC D10
^{125}I	1.00	1.00	1.00
^{111}In	0.55	0.67	0.74
^{177}Lu	0.49	0.62	0.71
^{131}I	0.32	0.46	0.57
^{90}Y	0.04	0.07	0.12

338

339 As radionuclides release more energy per decay, the penetration of the electrons / β^- particles
 340 increase (see Table 1 and Table 2). Thus, more energy is deposited outside the MALCs, and in turn the

341 absorbed fractions decrease. ^{125}I deposits nearly its whole electron energy within the three MALCs. The
 342 electrons it emits are unlikely to deposit energy outside the MALCs since their penetrations are limited
 343 to $1.5 \cdot 10^{-2}$ mm while the rituximab is mostly located at more than $\sim 3 \cdot 10^{-2}$ mm from the edge. Note that
 344 the real values are inferior to 0.998 (rounded to 1.00 for consistency), due to scarce rituximab biodis-
 345 tribution closer to or at the surface. The absorbed fractions for ^{111}In , lying between 0.55 and 0.74, are
 346 notably lower than that of ^{125}I . The reason is that $\sim 80\%$ of the electron energy released per decay of
 347 ^{111}In is taken away by emissions of at least 145.0 keV, characterized by penetrations above $1.8 \cdot 10^{-1}$ mm
 348 (right-hand side of Figure 3). Due to this property, the absorbed fractions for ^{111}In are remarkably close
 349 to ^{177}Lu , as their emissions have similar penetrations. Furthermore, the absorbed fractions for ^{131}I are
 350 higher than for ^{177}Lu as its penetration are higher (by $1.0 \cdot 10^{-1}$ mm at $E_{\beta^- \text{ mean}}$). Finally, the absorbed
 351 fractions for ^{90}Y , ranging between 0.04 and 0.12, are by far the lowest. Such low values are due to the
 352 penetrations of its emissions (up to 8.4 mm) being significantly superior to the size of the MALCs.

353 Figure 7 displays the mean absorbed dose per decay delivered to the MALCs by each radionu-
 354 clide and Table 5 shows the corresponding numerical values. These results are impacted by the energy
 355 released by the radionuclides and the absorbed fractions.



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Figure 7. Mean absorbed dose per decay in MALC D2, D5 and D10 for ^{125}I , ^{111}In , ^{177}Lu , ^{131}I and ^{90}Y .

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Table 5. Mean absorbed dose per decay in the MALCs for the five radionuclides.

Radionuclide	Mean absorbed dose per decay ($\mu\text{Gy} \cdot \text{Bq}^{-1} \cdot \text{s}^{-1}$)		
	MALC D2	MALC D5	MALC D10
^{125}I	$6.78 \cdot 10^{-2}$	$5.69 \cdot 10^{-3}$	$1.10 \cdot 10^{-3}$
^{111}In	$6.26 \cdot 10^{-2}$	$6.41 \cdot 10^{-3}$	$1.37 \cdot 10^{-3}$
^{177}Lu	$2.27 \cdot 10^{-1}$	$2.41 \cdot 10^{-2}$	$5.34 \cdot 10^{-3}$
^{131}I	$2.01 \cdot 10^{-1}$	$2.46 \cdot 10^{-2}$	$5.84 \cdot 10^{-3}$
^{90}Y	$1.21 \cdot 10^{-1}$	$1.98 \cdot 10^{-2}$	$6.15 \cdot 10^{-3}$

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Despite ^{111}In (34.7 keV/decay) releasing more energy than ^{125}I (19.5 keV/decay), the former delivered a lower absorbed dose in MALC D2 due to its lower absorbed fraction. Nevertheless, for MALC D5 and D10, the absorbed fraction for ^{111}In increases, leading to greater absorbed doses than for ^{125}I .

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In each MALC, absorbed doses for β^- particle emitters are greater than for Auger electron emitters, because they release considerably more energy. For example, absorbed doses for ^{177}Lu are greater than that of ^{111}In by a factor ~ 4 . Both have similar absorbed fractions, but ^{177}Lu (147.8 keV/decay) releases ~ 4 times more energy. For β^- particle emitters, ^{177}Lu and ^{131}I (191.7 keV/decay) results are close. Lower energy released by ^{177}Lu , is compensated by its higher absorbed fractions. Among β^- particle emitters, ^{90}Y (933.2 keV/decay) emits by far the more energy, yet the absorbed doses delivered are lowest in MALCs D2 and D5 due to extremely low absorbed fractions. In MALC D10, however, it delivered the highest absorbed doses because, compared to MALC D2, absorbed fractions were increased by a factor 3.

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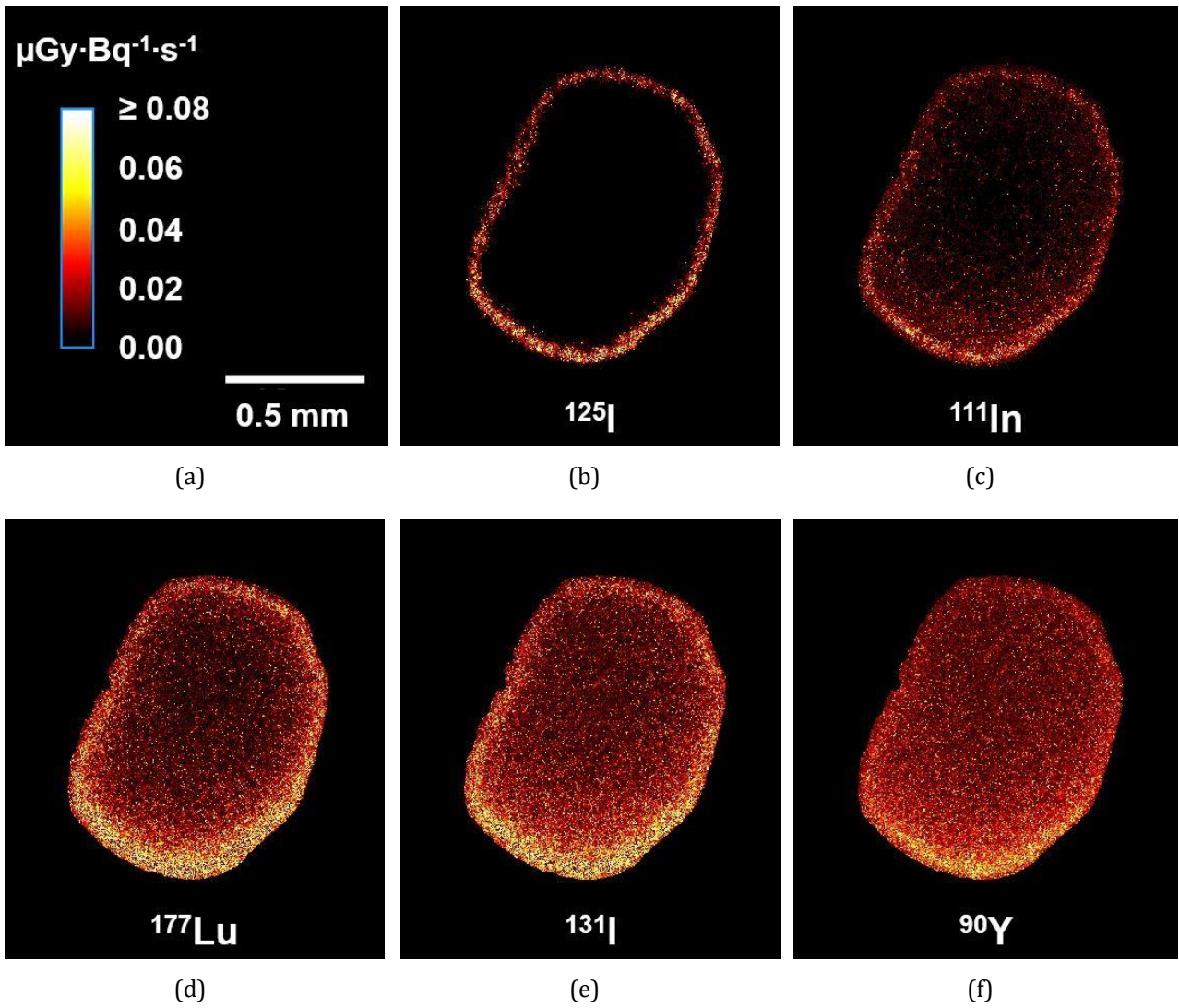
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Figure 8 illustrates, as an example the distributions of the absorbed dose per decay (in $\mu\text{Gy} \cdot \text{Bq}^{-1} \cdot \text{s}^{-1}$) inside the voxels of the slice of MALC D5 shown in Figure 1 (b). Note that $0.01 \mu\text{Gy} \cdot \text{Bq}^{-1} \cdot \text{s}^{-1}$ corresponds to a mean energy deposited per decay inside a voxel of $2.6 \cdot 10^{-3} \text{ eV} \cdot \text{Bq}^{-1} \cdot \text{s}^{-1}$. Figure 9 displays the cDVH (in $\mu\text{Gy} \cdot \text{Bq}^{-1} \cdot \text{s}^{-1}$) of MALCs (a) D2, (b) D5 and (c) D10 for ^{125}I , ^{111}In , ^{177}Lu , ^{131}I and ^{90}Y . Globally, when radionuclides release more energy, the absorbed dose is delivered more uniformly within the MALCs because of the increasing penetration of the emission.

380 Figure 8 (b) highlights the peripheral irradiation of ^{125}I , with a pattern very similar to the ritux-
381 imab biodistribution. However, this radionuclide delivered high absorbed dose as revealed by the less-
382 steep gradient of its cDVH. These observations are consistent with ^{125}I emitting low energy electrons,
383 which deposit most of the energy in a small distance from the decay site. The highly localized absorbed
384 dose delivered by ^{125}I was further observed at the cellular level by Arnaud *et al.*¹⁴. Unlike this radionu-
385 clide, ^{111}In sparsely delivered absorbed dose in the central part of the MALCs (Figure 8 (c)) due to the
386 higher penetrations of its emission. For both Auger electron emitters, the low absorbed dose delivered
387 in the central part of the MALCs (or the lack thereof) is linked to the rituximab peripheral biodistribu-
388 tion. Nevertheless, the limited penetration of rituximab inside the MALCs can be counterbalanced by a
389 radiolabeling with one of the β^- particle emitters.

390 This benefit, illustrated by Figure 8 (d), (e) and (f), was brought by the higher penetration of
391 their emission. It translates to higher cDVH values on the left-hand side (Figure 9). Among β^- particle
392 emitters, ^{90}Y energy deposition was the most uniform throughout the MALCs. In comparison, ^{177}Lu and
393 ^{131}I deposited less energy in the central part but more in the periphery. The improved absorbed dose
394 uniformity of ^{90}Y is further expressed by steeper gradients in its cDVH. It is linked to the variation of
395 penetration and stopping power of β^- particles with their energy. β^- particles emitted by ^{90}Y having
396 higher penetrations than for ^{177}Lu and ^{131}I , were more likely to reach the center of the MALCs. This dif-
397 ference is more pronounced as MALCs get bigger. Compared to ^{177}Lu and ^{131}I , emissions of ^{90}Y have lower
398 stopping powers (according to NIST database⁶¹, at $E_{\beta^- \text{ mean}}$: 0.34, 0.29 and 0.19 keV/ μm , respectively),
399 thus they deposited less energy close to the decay site.



400 Figure 8. Absorbed dose per decay ($\mu\text{Gy}\cdot\text{Bq}^{-1}\cdot\text{s}^{-1}$) delivered to the voxel of the slice of MALC D5 shown in Figure 1
 401 (b), for (b) ^{125}I , (c) ^{111}In , (d) ^{177}Lu , (e) ^{131}I and (f) ^{90}Y . The corresponding color scale is displayed in (a).

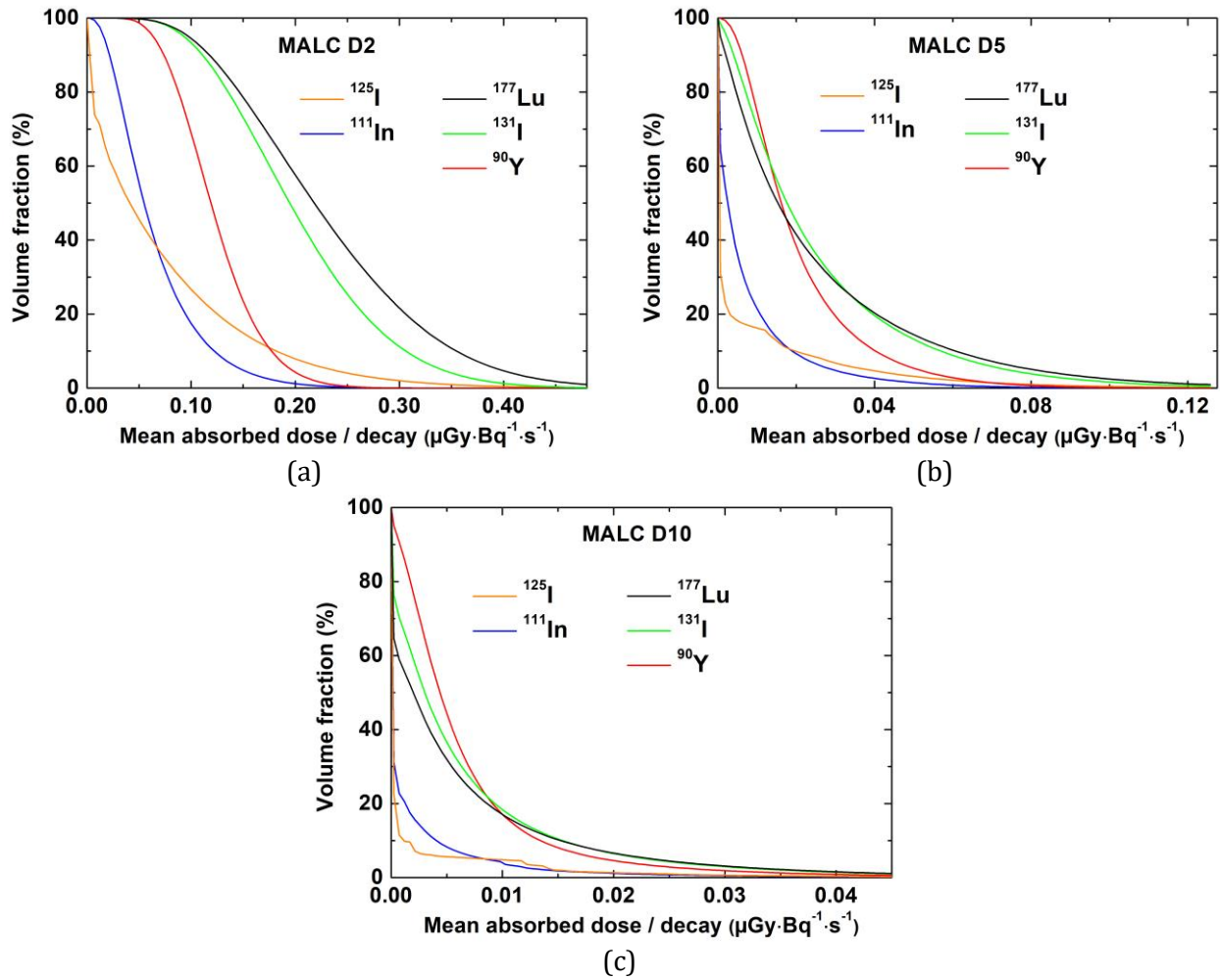


Figure 9. cDVH of MALC (a) D2, (b) D5 and (c) D10 per decay for ^{125}I , ^{111}In , ^{177}Lu , ^{131}I and ^{90}Y .

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404 3.3 Absorbed dose per unit administered activity

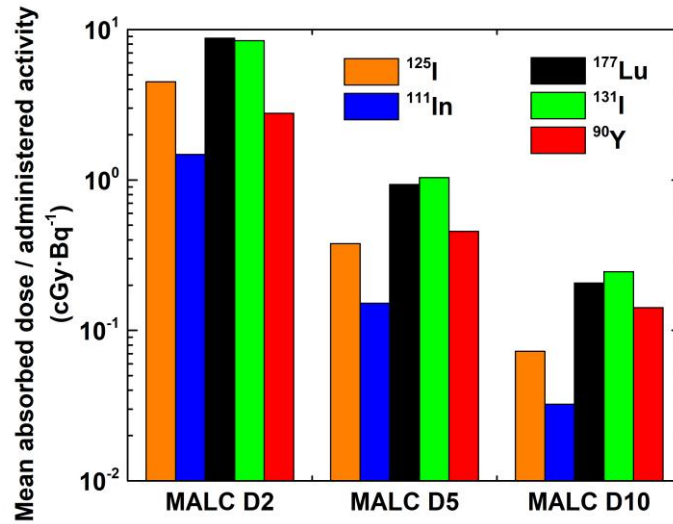
405 The influence of the radiolabeled rituximab effective half-life on the absorbed dose is presented.
406 Table 6 summarizes the effective half-lives for radiolabeling with ^{125}I , ^{111}In , ^{177}Lu , ^{131}I or ^{90}Y as calculated
407 from equation (6). The differences between the physical half-lives of the radionuclides were decreased
408 by the introduction of the biological half-life of the rituximab (5.8 days).

409 Table 6. Effective half-lives of rituximab radiolabeled with one of the five radionuclides in NHL, determined with
 410 equation (6).

Radionuclide X	Effective half-life of X-rituximab (day)
¹²⁵ I	5.3
¹¹¹ In	1.9
¹⁷⁷ Lu	3.1
¹³¹ I	3.4
⁹⁰ Y	1.8

411

412 For each radionuclide, Figure 10 displays the mean absorbed dose per unit administered activity
 413 within the MALCs and Table 7 shows the corresponding numerical values. Regarding Auger electron
 414 emitters, the effective half-life of ¹²⁵I-rituximab was 2.8 times higher than ¹¹¹In-rituximab. Therefore,
 415 despite lower absorbed dose delivered per decay in MALC D5 and D10, ¹²⁵I delivered more absorbed
 416 dose per unit administered activity than ¹¹¹In. For β^- particle emitters, the effective half-life of ¹⁷⁷Lu-
 417 rituximab and ¹³¹I-rituximab were 1.7 and 1.9 times greater than that of ⁹⁰Y-rituximab. As a result, ab-
 418 sorbed dose per unit administered activity for ¹⁷⁷Lu and ¹³¹I were more than 3 times higher in MALC D2
 419 and more than 2 times higher in MALC D5 than that of ⁹⁰Y. Furthermore, in MALC D10, despite higher
 420 absorbed dose per decay, ⁹⁰Y delivered less absorbed dose per unit administered activity than ¹⁷⁷Lu and
 421 ¹³¹I. As discussed by Scheidhauer *et al.*⁵⁸, the relatively long biological half-life of rituximab (5.8 days in
 422 NHL) is best matched by ¹³¹I physical half-life (8.0 days) than by ⁹⁰Y short physical half-life (2.7 days).
 423 ¹⁷⁷Lu physical half-life of 6.6 days is also well adapted to the rituximab biological half-life.



424

425 Figure 10. Mean absorbed dose per unit administered activity to MALCs D2, D5 and D10 for ¹²⁵I, ¹¹¹In, ¹⁷⁷Lu, ¹³¹I
 426 and ⁹⁰Y.

427

428

Table 7. Mean absorbed dose per unit administered activity to the MALCs for the five radionuclides.

Radionuclide	Mean absorbed dose per unit administered activity (cGy · Bq ⁻¹)		
	MALC D2	MALC D5	MALC D10
¹²⁵ I	4.50	3.78 · 10 ⁻¹	7.27 · 10 ⁻²
¹¹¹ In	1.48	1.51 · 10 ⁻¹	3.23 · 10 ⁻²
¹⁷⁷ Lu	8.76	9.33 · 10 ⁻¹	2.06 · 10 ⁻¹
¹³¹ I	8.45	1.04	2.46 · 10 ⁻¹
⁹⁰ Y	2.78	4.56 · 10 ⁻¹	1.42 · 10 ⁻¹

429

430 Figure 11 illustrates, as an example the isodoses (in cGy·Bq⁻¹) inside the voxels of the slice of
 431 MALC D5 shown by Figure 1 (b). Only the β- particle emitters were represented since the absorbed dose
 432 distribution of Auger electron emitters were very heterogeneous and did not allow for a clear represen-
 433 tation of isodoses. Figure 12 shows the cDVH of MALC (a) D2, (b) D5 and (c) D10 per unit administered
 434 activity for ¹²⁵I, ¹¹¹In, ¹⁷⁷Lu, ¹³¹I and ⁹⁰Y.

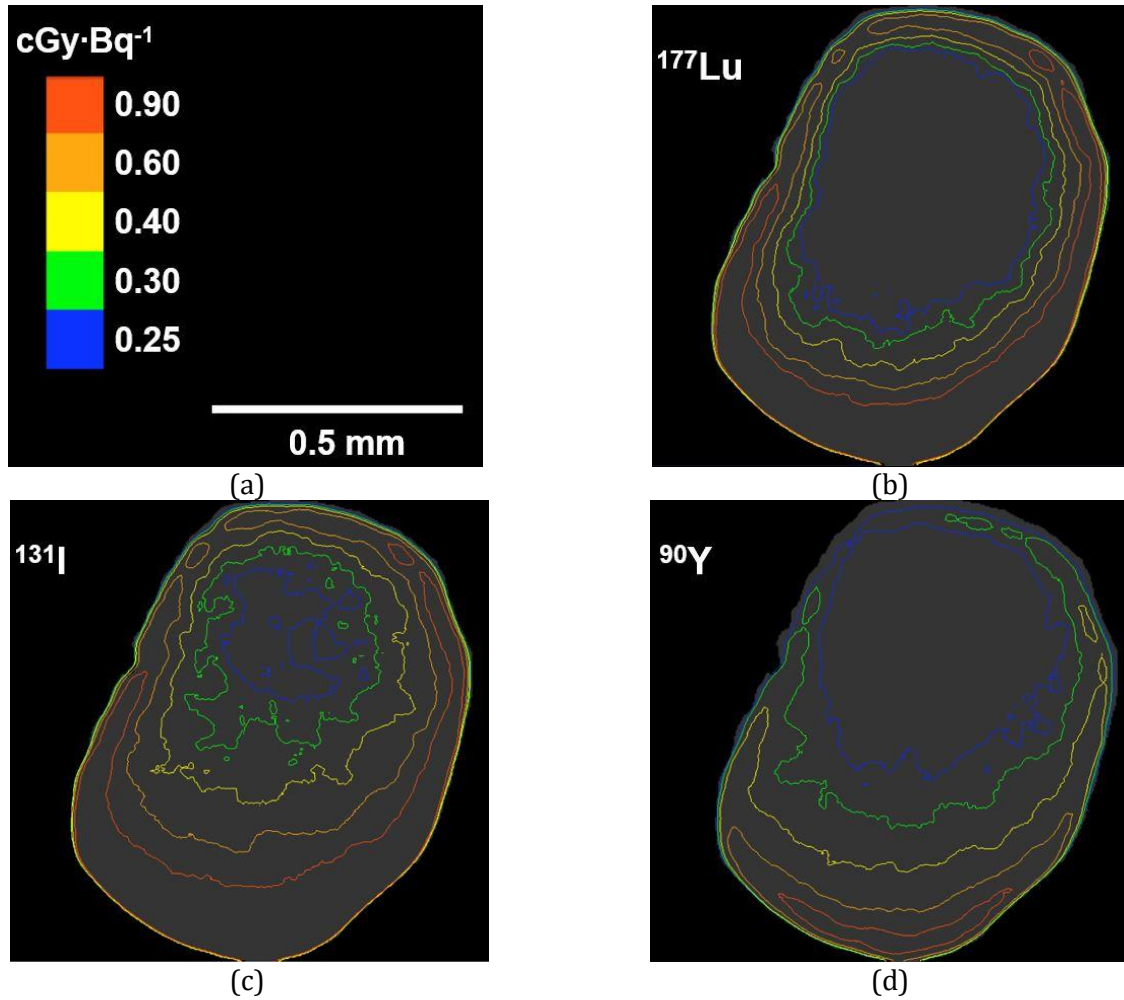
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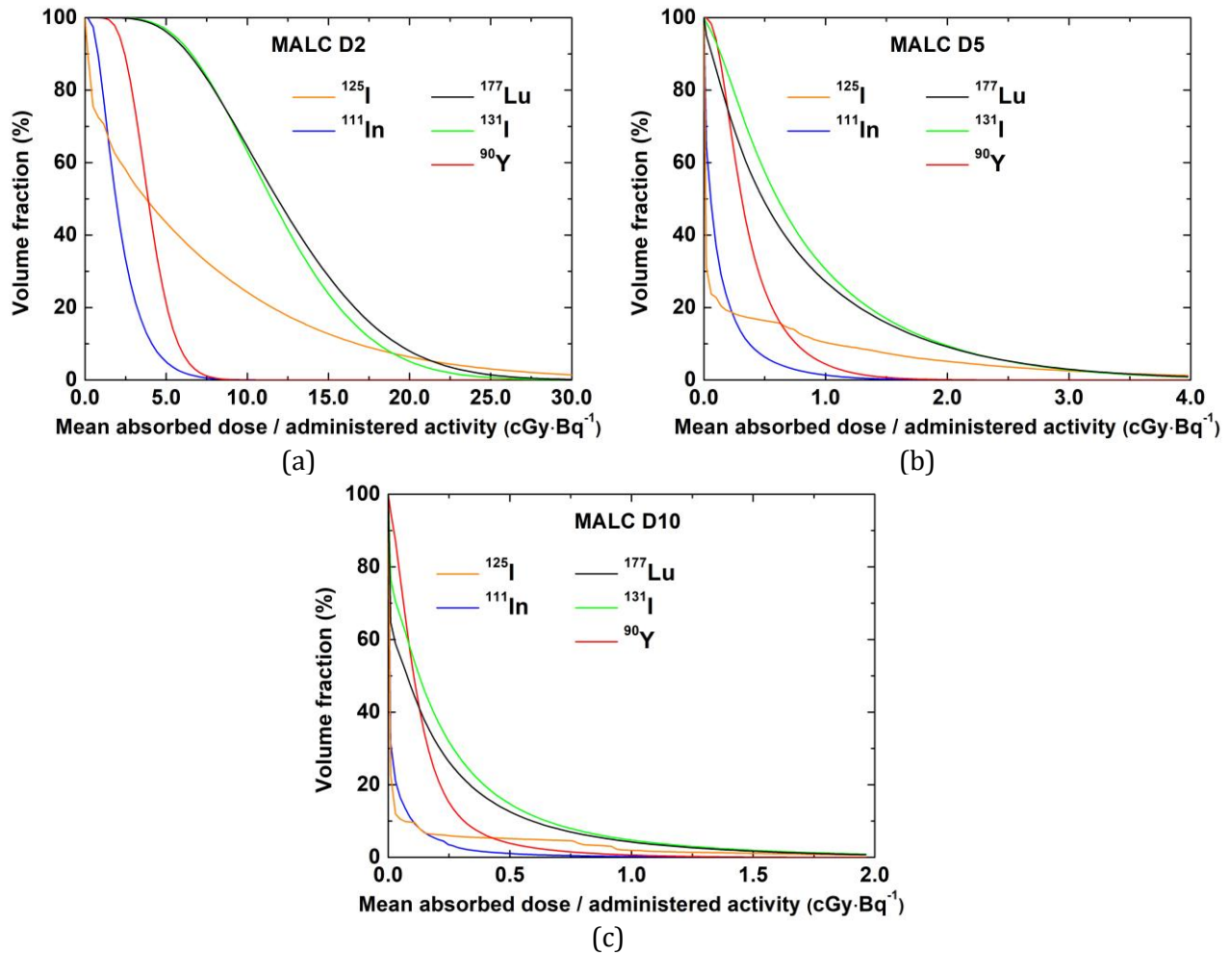
When the effective half-life of radiolabeled rituximab increases, the absorbed doses per admin-
 istered activity increase. As a consequence, the region covered by high absorbed dose is largest for ¹³¹I
 and smallest for ⁹⁰Y as illustrated by the isodoses (Figure 11). Furthermore, the higher the effective half-

438 life, the greater the shift of the cDVH curves toward higher absorbed doses. Accordingly, in the region of
 439 high absorbed doses, cDVH values of ^{125}I become significantly higher than that of ^{111}In . In this region,
 440 cDVH values of ^{177}Lu and ^{131}I are noticeably higher than those of ^{90}Y .



441 Figure 11. Isodose distributions in the slice of MALC D5 shown in Figure 1 (b), per unit administered activity of
 442 (b) ^{177}Lu , (c) ^{131}I and (d) ^{90}Y (cGy·Bq⁻¹). The corresponding color scale is displayed in (a).

443



444 Figure 12. cDVH per unit administered activity of MALC (a) D2, (b) D5 and (c) D10 for ^{125}I , ^{111}In , ^{177}Lu , ^{131}I and
 445 ^{90}Y .

446 3.4 Discussion

447 Our methodology allowed the observation of the absorbed dose in an *in vitro* tumor model which
 448 is more realistic than the spherical tumor with uniform biodistribution simplification. It is recognized,
 449 however, that the realism of this work is limited with respect to clinical context. It is therefore essential
 450 to review the degree of relevance of the biological and physical aspects of this work.

451 The present study focused on three MALCs with dimensions of the order of millimeter. On one
 452 hand, such a sample did not reproduce the diversity of tumor geometry and rituximab pharmacokinetics
 453 in patients, which are critical for dosimetry. For example, in clinical routine, the size of the diagnosed
 454 follicular lymphomas ranges from a few millimeters to several centimeters⁶². Additionally, multiple
 455 studies have highlighted the need for dosimetry in smaller follicular lymphomas (down to $\sim 10\ \mu\text{m}$)⁶³⁻⁶⁵.
 456 Our study did not extend to the biggest and the smallest of the follicular lymphomas. It is expected that,
 457 with the increasing / decreasing size of the tumor model, radionuclides with higher (*e.g.* ^{90}Y) / lower

458 (e.g. ^{125}I) penetration emissions, would become more favored from a dosimetric perspective. On the
459 other hand, previous studies have highlighted the ability of three-dimensional aggregates to mimic the
460 physiological *in vivo* aspects of tumors, enabling the identification of new targets and drugs⁶⁶⁻⁶⁸. In par-
461 ticular, this tool has long been recognized as very effective to characterize the penetration of antibod-
462 ies⁶⁹⁻⁷⁰. These previous works give confidence in the clinical relevance of the rituximab peripheral dis-
463 tribution in the MALCs, a key point of the present study.

464 The accurate Monte Carlo tracking of the radionuclide emissions in the MALCs relied largely on
465 the choice of the physics models. Geant4-DNA provides two sets of physics models able to entirely sim-
466 ulate the electron emissions of ^{125}I and ^{111}In : the default (“option 2”) and the CPA100 (“option 6”) sets.
467 The latter set was selected following two main arguments. Differential cross sections for ionization and
468 cross section for electronic excitation of “option 6” are in better agreement than that of “option 2” with
469 experimental data. Furthermore, for low energy electrons, dose-point kernels generated by “option 2”
470 are considered unrealistic because they are very diffusive⁷¹, while those calculated by “option 6” are less
471 so⁷². Concerning β^- particle emitters, “Livermore” were considered more relevant than “PENELOPE-
472 2008”⁷³ electron physics models based on two arguments. Firstly, to be in accordance with Geant4-DNA
473 “option 6”, the simulations were realized in liquid water. For this material, “Livermore” ionization cross
474 sections are in better agreement with experimental data³⁶. Secondly, regarding deexcitation processes,
475 “Livermore” simulates every atomic shell, while “PENELOPE-2008” only takes into account K, L and M
476 shells⁷³.

477 Monte Carlo simulations were informed with data from one SPIM acquisition, showing biodis-
478 tribution in MALCs after 24 h of incubation. To approximate the time-dependent activity in the MALCs,
479 we used the straightforward assumptions of a radiopharmaceutical instantaneous uptake and monoex-
480 ponential elimination. In clinical studies, a more detailed description of the uptake and elimination is
481 provided. Typically, 3 to 4 SPECT or PET acquisitions are performed, the first a few hours and the last
482 between a few tenth of hours and nearly 200 h after administration^{25,56,74-75}. Generally, these data are
483 fitted by a mono/biexponential function. The instantaneous uptake does not account for the decays oc-
484 ccurring outside the tumor during the uptake in a clinical context. Hence, the absorbed dose to the tumor

485 delivered by radionuclides with short half-lives, such as ^{111}In and ^{90}Y , is overestimated compared to
486 those with longer half-life. We note, however, that SPECT acquisitions of mice and patients showed early
487 uptake of rituximab in NHL⁷⁶⁻⁷⁷ (a few hours after administration while ^{111}In and ^{90}Y physical half-lives
488 are ~ 2 days). Regarding the elimination of the rituximab in NHL, no data were found in the literature.
489 Implementation of more sophisticated pharmacokinetic models in our methodology would be an im-
490 portant next step for a more relevant dosimetry. Ideally, these models should consider rituximab data
491 in patients' NHL, upon availability.

492 **4 Conclusion**

493 We have developed a Monte Carlo dosimetric model based on realistic NHL geometry and ritux-
494 imab biodistribution. It was employed to describe the absorbed dose that would be delivered at the
495 microscopic scale by Auger electron emitters (^{125}I and ^{111}In) and β^- particle emitters (^{177}Lu , ^{131}I and ^{90}Y)
496 in the context of RIT.

497 A comparison between the mean absorbed dose per decay for realistic geometry and biodistri-
498 bution with literature data for a sphere with either a surface or a volume biodistribution demonstrated
499 significant differences. Then, for the five radionuclides, the absorbed fractions, the mean absorbed dose
500 and its associated distribution were compared. This comparison further highlighted the importance of
501 both the realistic MALC geometry and rituximab biodistribution as they often impacted the outcomes.

502 Aside from these two parameters, the decay properties - the electron / β^- particle energy re-
503 leased per decay and penetration - had an important impact. The uniformity of the absorbed dose deliv-
504 ered throughout the MALCs increased for radionuclides releasing more energy per decay (in ascending
505 order: ^{125}I , ^{111}In , ^{177}Lu , ^{131}I and ^{90}Y). The mean absorbed dose of β^- particle emitters was higher than that
506 of Auger electron emitters. ^{177}Lu and ^{131}I mean absorbed doses and were higher than ^{90}Y in the two
507 smallest MALCs. Furthermore, when considering the radiolabeled rituximab effective half-life, the ab-
508 sorbed dose for ^{177}Lu and ^{131}I remained close to each other and considerably greater than that of ^{90}Y in
509 the three MALCs.

510 The methodology developed may be applied to various situations of TRT. Voxelized images of
511 others tumoral / healthy tissue models and tumor-seeking agents can be considered. It is possible to

512 simulate the emission of other radionuclides of interest such as potential theranostic Auger electron
 513 emitters (*e.g.* copper-64⁷⁴, useful for PET imaging) and β^- particle emitters (*e.g.* terbium-161⁷⁸⁻⁷⁹, useful
 514 for SPECT imaging) or therapeutic α particle emitters (*e.g.* actinium-225⁸⁰⁻⁸¹). To further improve the
 515 realism of the simulations, future work may integrate the temporal variation of the vector biodistribu-
 516 tion and the growth of MALC. Such work could be useful in the case of real irradiation experiment to
 517 select relevant radionuclides and assess biological effects.

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744 **6 Acknowledgments**

745 We would like to express our gratitude to Prof. Steve Archibald (PET Research Centre, University
746 of Hull) and Dr Jean-Luc Gerner (ESA/ESTEC, retired) for helpful advice. We thank Lise Teyssedre
747 (ITAV/CNRS) for fruitful discussion of SPIM imaging.

748 **7 Conflict of interest statement**

749 The authors declare that there is no conflict of interest regarding the publication of this article.