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# Female patient-specific finite element modeling of pelvic organ prolapse (POP)

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## Abstract

Pelvic organ prolapse (POP) occurs only in women and becomes more common as women age. However, the surgical practices remain poorly evaluated. The realization of a simulator of the dynamic behavior of the pelvic organs is then identified as a need. It allows the surgeon to estimate the functional impact of his actions before his implementation. In this work, the simulation will be based on a patient-specific approach in which each geometrical model will be carried out starting from magnetic resonance image (MRI) acquisition of pelvic organs of one patient. To determine the strain and stress in the biological soft tissues, hyperelastic constitutive laws are used in the context of finite element analysis. The Yeoh model has been implemented into an in-house finite element code FER to model these organ tissues taking into account large deformations with multiple contacts. A 2D model is considered in this

preliminary study and the results show that our method can help to improve the understanding of different forms of POP.

*Key words:* Pelvic organ prolapse, Finite element method, Soft tissue, Hyperelasticity, Deformation

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## 1 Introduction

Pelvic organ prolapse (POP) occurs only in women and becomes more common as women age. It refers to the loss of support to the structures contained within the bony pelvis and describes the descent of the pelvic organs into the vagina (Luft, 2006). By some estimates, about half of parous women experience some degree of prolapse and that 10-20% seeks medical care (Thakar & Stanton, 2002). POP repairs consist of 20% of the gynaecological surgeries (Jackson & Smith, 1997). Symptoms associated with POP are various degrees of pelvic pressure, low back pain, vulvovaginal excoriation, bowel dysfunction and sexual difficulties (Hagen, Stark, & Cattermole, 2004). Reported by US National Center for Health Statistics that in 2007, POP was a common diagnoses associated with hysterectomy, resulting in more than 517,000 surgical procedures (Hall, DeFrances, Williams, Golosinskiy, & Schwartzman, 2010). However, the surgical practices remain poorly evaluated. It has been estimated that the recurrence of POP in nearly one third of women after repair surgery (Olsen, Smith, Bergstrom, Colling, & Clark, 1997), and that 80% of these cases recur within two years of surgery (Baden & Walker, 1992). The use of precise

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18 numerical models of the female pelvic system will, in the future, provide the  
19 tools to simulate the dynamic behavior of pelvic organs. The realization of  
20 a simulator will allow surgeons to estimate the functional impact of his ac-  
21 tions before implementation, to perform the surgery in a more controlled and  
22 reliable way.

23 In order to achieve these goals, we need to study the morphology and anatomy  
24 of these organs, to understand their environment and function to obtain mean-  
25 ingful boundary conditions and to determine material properties (Fung, 1993).  
26 From a mechanical point of view, the pelvic organs have specific geometries  
27 and material properties. They can be thought of as biomechanical structures  
28 which move and interact with each other due to external pressures, while the  
29 constraints are induced by rigid bodies (bones) as well as soft tissues such as  
30 muscles, ligaments and fascias, all ensuring the stability of these organs.

31 In the modeling works of the consulted literatures, biological soft tissues are  
32 usually subjected to large deformations with negligible volume change and  
33 show a highly non-linear anisotropic mechanical response due to their internal  
34 structure (Pena, Martínez, Calvo, & Doblar, 2006). To determine the strain  
35 and stress in the biological soft tissues such as ligaments, tendons or arterial  
36 walls, Spencer (Spencer, 1954) firstly developed the framework of continuum  
37 mechanics by means of the definition of a strain energy density function ex-  
38 pressed in terms of kinematic invariants to model the purely elastic response of  
39 these tissues. This approach was further adapted to finite element formulations  
40 of soft collagenous biological tissues (see for example (Weiss, Maker, & Govin-  
41 djee, 1996; Pena, del Palomar, Calvo, Martínez, & Doblar, 2007) for ligaments  
42 and (Holzapfel, Gasser, & Ogden, 2000) for arteries). However, a large num-  
43 ber of elements of living organisms show very distinct anisotropic mechanical

44 properties (Holzapfel, 2006). Most energy densities used to model transversely  
45 isotropic and orthotropic soft tissues take a power law form (Schröder, Neff, &  
46 Balzani, 2005) or present an exponential behavior (Fung, Fronek, & Patitucci,  
47 1979; Holzapfel et al., 2000). Although there might be some distinctions be-  
48 tween quadrupedal and bipedal, the mechanical anisotropy properties of ewe  
49 vaginal tissue was highlighted in the tests of Rubod et al. (Rubod, Bouker-  
50 rou, Brieu, Dubois, & Cosson, 2006a). Uniaxial tension tests were performed  
51 on the samples of vaginal tissue of patients with POP and fresh cadavers  
52 without prolapse (nPOP) (Jean-Charles et al., 2010), it shows that POP is  
53 significantly stiffer than nPOP tissue, both on anterior and posterior walls.  
54 Biomechanical properties of the vagina, aponeurosis and skin were compared  
55 (Gabriel et al., 2011), it turns out that the aponeurosis was much more rigid  
56 and less extendible than the vagina and skin, vaginal tissue was less rigid but  
57 more extendible than skin. In our work, Yeoh is chosen to model the isotropic  
58 and hyperelastic behavior of pelvic organs, and different material parameters  
59 are considered for different parts of the pelvic system.

60 Numerical simulations of interactions between male pelvic organs (bladder,  
61 prostate and rectum) and modeling approaches of female POP have already  
62 been performed (Boubaker, Haboussi, Ganghoffer, & Aletti, 2009; Venugopala Rao,  
63 Rubod, Brieu, Bhatnagar, & Cosson, 2010). Since intra-abdominal pressure  
64 has been considered as the cause of some multi-organ dysfunction (Cobb et al.,  
65 2005; Chionh, Wei, Martin, & Opdam, 2006), a 2.5 MPa pressure is applied on  
66 the surface of pelvic organ models by Venugopala Rao et al. (Venugopala Rao  
67 et al., 2010) while in the work of Boubaker et al. (Boubaker et al., 2009), only  
68 internal pressures are applied on the internal surfaces of bladder and rectum.  
69 A ‘bear down’ simulation of pelvic floor muscles was performed by Noakes

70 et al. (Noakes, Pullan, Bissett, & Cheng, 2008). In their work, an average  
71 downwards displacement of 27.2 mm and an average posterior displacement  
72 of 18.4 mm were produced due to a pressure of 4 kPa applied on the levator  
73 ani. It's very cumbersome to estimate the exact load on the pelvic system, so  
74 in present work, a displacement of 27.9 mm is identified and applied on the  
75 digitized surface of the pelvic floor muscle in the dynamic magnetic resonance  
76 image (MRI) sequence (Noakes et al., 2008).

77 The purpose of this study is to present preliminary 2D simulation results of  
78 interactions between pelvic organs during prolapse process using the finite  
79 element method on computational meshes based on live subject data. The  
80 paper is organized as follows: in Section 2, the establishment of a geometrical  
81 model is introduced based on the MRI of a specific patient. In Section 3, the  
82 anatomic description of the pelvic system is introduced. The structural and  
83 mechanical properties of the organ tissues are presented. In Section 4, the  
84 material parameters of the postulated constitutive laws are identified from lit-  
85 eratures and the boundary conditions are established. Two simulation results  
86 are presented and discussed afterwards. In the last section, the conclusion and  
87 some perspectives of this project are proposed.

## 88 **2 Geometrical definition of pelvic organs**

89 To determine a realistic behavior of the organs, an accurate geometric model  
90 associated with a physical modeling is required. The geometric modeling of  
91 pelvic organs consists of medical imaging to reconstruct 3D models on which  
92 we must implement physical models and a numerical solution based on a  
93 spatial discretization from the finite element method. A geometrical rebuilding

94 of the pelvic organs of a female patient with no history of pelvic pathology  
95 perineal is performed with MRI data (Pirr6 et al., 2009). The acquisition of T2  
96 static sagittal images of pelvic organs, with a thickness of 1 mm, was obtained  
97 in the supine position using a surface antenna (TR = 5.5 ms, TE = 2.7 ms,  
98  $256 \times 256$  matrix) (Fig. 1). This acquisition was the basis of pelvic model. A  
99 sagittal dynamic acquisition of thick sections of 10 mm was also performed at  
100 rest and during thrust using a dynamic turbo T2 sequence (TR = 3.6 ms, TE  
101 = 1.8 ms, FOV = 24 cm,  $256 \times 256$  matrix).

102 The three-dimensional solid model was obtained from static images of MRI  
103 using free software segmentation (ITK-SNAP) based on the evolution of an  
104 active contour. The software can visualize simultaneously the organs in three  
105 planes and define the boundaries of areas to be segmented. A volumetric germ  
106 was placed in the defined area (Fig. 2), it grew automatically until it reached  
107 the defined limits. The cease of segmentation was automatic and was based  
108 on the variation of volume of the volumetric germ. At the end of the segmen-  
109 tation process, the geometrical irregularities were automatically corrected by  
110 smoothing, the three dimensional geometric model was generated as shown  
111 in Fig. 3. A thickness is added with an offset formulation, and the geometric  
112 model is finally exported in a hexahedral mesh (Bay et al., 2012). As a pre-  
113 liminary study of pelvic organs, a 2D model obtained in the sagittal plane is  
114 considered to test the capability of the developed algorithms and software to  
115 deal with large deformation hyperelastic problems with contact interfaces.

### 116 **3 Characterization of pelvic organs**

117 To perform a numerical simulation of the pelvic organ's motion and deforma-  
118 tion, it is essential to have a good recognition of the environment and location  
119 of the pelvic organs and to determine their material properties.

#### 120 *3.1 From an anatomical point of view*

121 To have a good geometrical description of the true pelvis also called pelvic  
122 cavity leads us to the recognition of the anatomy of the pelvic organs (bladder,  
123 uterus and rectum). Fig. 4 illustrates a normal female pelvic anatomy. The  
124 pelvic floor lies at the bottom of the abdominal cavity, with this position, it  
125 acts as a supportive layer that prevents the pelvic organs from falling down  
126 (DeLancey, 1993; Messelink et al., 2005). Bladder and rectum are positioned  
127 anteriorly and posteriorly in the pelvic cavity. The bladder is supported by the  
128 adjacent parts of the skeleton and the pelvic floor (Drake, Vogl, & Mitchell,  
129 2010). The urethra passes through the pelvic floor to reach the perineum,  
130 where it opens outside. Following the sigmoid colon at the vertebra, the rectum  
131 ends in the anal canal, which crosses the pelvic floor to open in the perineum.  
132 The pelvic cavity contains most of the reproductive tract of women. The vagina  
133 crosses the pelvic floor and is inserted into the uterus in the pelvic cavity which  
134 is placed between the rectum and bladder (Drake et al., 2010). The uterus is  
135 stabilized by the existence of transverse ligaments, anterior (pubocervical)  
136 and posterior (utero-sacral) formed by the pelvic fascia and connected to the  
137 pelvis walls. These ligaments would act primarily at the cervix leaving some  
138 rotational mobility of the uterus to adapt to the repletion of the bladder and



139 rectum.

140 According to the anatomical analysis, the existing ligaments in a pelvic system  
141 for fixation are difficult to precisely individualise on MRI, thus we have these  
142 following biomechanical considerations for the mobility of organs:

- 143 • The outlet orifices of the urethra and vagina in the perineum are almost fixed  
144 points relative to pubic bone (except serious malfunction of the perineum).
- 145 • The musculoskeletal pelvis wall can be considered as fixed and rigid with  
146 respect to the elasticity of the pelvic organs.
- 147 • The rectum is attached to the pelvic floor. The dorsal part of the rectum  
148 almost does not move, it relies on the back of the pelvic floor.
- 149 • The bladder is attached to a part of the fascia stretched along a plane  
150 of inclination at the point of connection of two ureters and the urethra,  
151 acting like a hammock. It is slightly held at its apex in the direction of the  
152 umbilicus. It is also surrounded by lateral ligaments belonging to the fascia,  
153 which start from the pubis to rejoin the cervix.
- 154 • The uterus is supported at the cervix by a set of ligaments from the fascia.  
155 The vagina is maintained by the fascia more or less rigidly following the  
156 stiffness of this part of the fascia. The vagina is kept separate from the  
157 rectum through the septum.

### 158 *3.2 Mechanical properties and constitutive laws*

159 To perform the numerical simulation, it is essential to characterize the mechan-  
160 ical behavior of pelvic tissues. Many uniaxial tension tests of pelvic tissues were  
161 performed and stress-strain curves were obtained (Rubod, Boukerrou, Brieu,

162 Dubois, & Cosson, 2006b; Gabriel et al., 2011) and a hyperelastic behavior  
 163 of the tissues was highlighted. To determine the strain and stress in the bio-  
 164 logical soft tissues such as ligaments, tendons, pelvic organs or arterial walls,  
 165 hyperelastic constitutive laws are often used in the context of finite element  
 166 analysis. Many attempts have been made to develop a theoretical stress-strain  
 167 relation that fits experimental results for hyperelastic materials (Charlton &  
 168 Yang, 1994; Boyce & Arruda, 2000; Miehe, Göktepe, & Lulei, 2004).

169 In order to describe the geometrical transformation problems, the deformation  
 170 gradient tensor is defined by

$$\mathbf{F} = \mathbf{I} + \nabla \mathbf{u}, \quad J = \det(\mathbf{F}) > 0 \quad (1)$$

171 where  $\mathbf{I}$  is the unity tensor and  $\mathbf{u}$  the displacement vector. Because of large  
 172 displacements and rotations, Green-Lagrangian strain is adopted for the non-  
 173 linear relationships between strains and displacements. We note  $\mathbf{C}$  the stretch  
 174 tensor or the right Cauchy-Green deformation tensor ( $\mathbf{C} = \mathbf{F}^T \mathbf{F}$ ). The Green-  
 175 Lagrangian strain tensor  $\mathbf{E}$  is defined by

$$\mathbf{E} = (\mathbf{C} - \mathbf{I})/2 \quad (2)$$

176 In the case of hyperelastic law, there exists an elastic potential function  $W$  (or  
 177 strain energy density function) which is a scale function of one of the strain  
 178 tensors, whose derivative with respect to a strain component determines the  
 179 corresponding stress component. This can be expressed by

$$\mathbf{S} = \frac{\partial W}{\partial \mathbf{E}} = 2 \frac{\partial W}{\partial \mathbf{C}} \quad (3)$$

180 where  $\mathbf{S}$  is the second Piola-Kirchoff stress tensor. In the particular case of  
 181 isotropic hyperelasticity, Eq. (3) can be written by

$$\mathbf{S} = 2 \left[ I_3 \frac{\partial W}{\partial I_3} \mathbf{C}^{-1} + \left( \frac{\partial W}{\partial I_1} + I_1 \frac{\partial W}{\partial I_2} \right) \mathbf{I} - \frac{\partial W}{\partial I_2} \mathbf{C} \right] \quad (4)$$

182 where  $I_i$  ( $i = 1, 2, 3$ ) denote the invariants of the right Cauchy-Green deforma-  
 183 tion tensor  $\mathbf{C}$  :

$$I_1 = tr(\mathbf{C}) ; I_2 = (I_1^2 - tr(\mathbf{C}^2))/2 ; I_3 = det(\mathbf{C}) \quad (5)$$

184 Among the various energy density functions available in the literature (Mooney,  
 185 1940; Rivlin, 1948; Arruda & Boyce, 1993; Harth-Smith, 1996; Gent, 1996;  
 186 Lambert-Diani & Rey, 1998; Feng, Vallée, Fortuné, & Peyraut, 2006), we have  
 187 chosen the Yeoh constitutive law for the simulation of the isotropic soft tissues.  
 188 This model leads to a mathematical function that fits well with the experi-  
 189 mental data because it takes into account the global form of the experimental  
 190 stress-strain curve. This model has been implemented into the in-house finite  
 191 element code FER (Feng, 2008). The Yeoh strain energy density formulation  
 192 is given by

$$W(I_1) = \sum_{i=1}^3 C_{i0}(I_1 - 3)^i + \frac{K}{2}(J - 1)^2 \quad (6)$$

193 where  $C_{i0}$  are the material constants.  $K$  is user-defined parameter which can  
 194 be interpreted as the bulk modulus of the material. It is noted that if  $J$   
 195 approaches to unity, the last term of Eq. (6) will approach to zero, which  
 196 corresponds to an incompressible Yeoh material.

197 By deriving the energy density Eq. (6) with respect to the three invariants,  
 198 then reporting the results in Eq. (4) gives

$$\mathbf{S} = 2\left(\sum_{i=1}^3 iC_{i0}(I_1 - 3)^{i-1}\right)\mathbf{I} + K(J^2 - J)\mathbf{C}^{-1} \quad (7)$$

#### 199 **4 Finite element modeling**

200 In order to test our algorithms on solving of hyperelastic problems with con-  
 201 tact, we considered in our present work to simulate directly a 2D model in the  
 202 sagittal plane. In this case, we start by selecting nodes belonging to a sagittal  
 203 plane; then from these selected points, we create the contours in the plane  
 204 using spline functions, which allows us to define the surfaces to be meshed.

205 This 2D model includes 3262 plane quadrilateral elements and 3430 nodes  
 206 (Fig. 5). The organs were meshed as filled but with different materials zones  
 207 shown with different colors as depicted in Fig. 5, in which orange represents the  
 208 bladder, green represents the uterus and vagina, blue represents the rectum.  
 209 The yellow area represents the nearly incompressible material which can be

210 assumed as liquid inside these organs, the effect of shear on the material  
211 surrounding is neglected. The fascia connecting organs is represented by dark  
212 green area. The pubis is next to the bladder and is considered as fixed.

213 As introduced previously in Section 3, the pelvic organs have hyperelastic ma-  
214 terial properties. We decided to identify the coefficients of Yeoh model for each  
215 organ from the experimental results of Venugopala Rao et al. (Venugopala Rao  
216 et al., 2010), because the results are consistent with those of Boubaker’s thesis  
217 (Boubaker, 2009), by using the identification function of MSC.Marc Mentat  
218 software. The material parameters for each organ from experimental data are  
219 obtained and presented in Table 1. The pubis is fixed without considering its  
220 deformation. To the authors’ knowledge, there is no reference for the material  
221 inside the organs and the fascia, so we chose the parameters which make them  
222 more soft and incompressible than organs.

223 Two simulations of the 2D model are performed. The first is the prolapse of  
224 pelvic organs due to a single displacement, the second is a direct comparison  
225 with dynamic MRI images taken in the sagittal plane. The finite element sim-  
226 ulation of interactions between pelvic organs is performed in the framework of  
227 large deformation, the total Lagrangian formulation is selected in this work to  
228 describe nonlinear behavior. A frictionless sliding contact with no interpen-  
229 tration is considered between the deformable bodies (uterus and bladder) and  
230 between the bladder and pubis (Feng, Peyraut, & Labed, 2003; Feng, Peyraut,  
231 & He, 2006).

232 *4.1 Simulation based on data from the literature*

233 It is difficult to estimate the exact load imposed on pelvic organs, therefore, for  
234 the first simulation, an inferior displacement of 27.9 mm which was identified  
235 on the digitized surface of the pelvic floor muscle in the dynamic MRI image  
236 sequence (Noakes et al., 2008) is applied on the upper anterior part of uterus  
237 to see the interactions and deformations of the pelvic system. The simulation  
238 was performed by the in-house solver FER and the results are displayed by  
239 the postprocessing software FER/View (Feng & Feng, 2004).

240 The boundary conditions are specified in terms of node displacements: the  
241 pubic bone is totally constrained ( $U_x = U_y = 0$ ). As mentioned before, the  
242 pelvic floor lies at the bottom of these organs and acts as a supportive layer,  
243 although it was not defined in our model, but its effect was considered equally  
244 by fixing in one direction of the bottom of bladder, vagina and rectum, while  
245 allowing a little decline during POP ( $U_x = 0, U_y = -10$  mm).

246 The simulation result is shown in Fig. 6. A comparison between initial con-  
247 figuration and deformed shape at last step is presented with the distribution  
248 of von Mises stress. We can see that the organ deformed as expected and the  
249 multi-contact between these organs was well handled. The maximum stress  
250 seems to occur in the anterior and posterior part of the vagina. It is well  
251 known that the anterior vaginal wall prolapse, clinically known as cystocele,  
252 is the most common form of pelvic organ prolapses (DeLancey, 1992). Among  
253 the women who had some form of prolapse, 34.3% have cystocele, 18.6% have  
254 rectocele and 14.2% have uterine prolapse (Hendrix et al., 2002). Both the  
255 cystocele and rectocele occur when the tissues and muscles that hold the or-

256 gans in place are stretched or weakened. They cause separately the bladder  
257 and rectum to move from its natural position and press against the forward  
258 and back wall of the vagina, as shown in the simulation result. Therefore, this  
259 model as well as our software could help understand the reasons of such form  
260 of prolapse and estimate the virtual effect of the organ's interactions.

#### 261 *4.2 Simulation based on dynamic MRI images*

262 The dynamic MRI acquisition, performed with a 1.5 T Philips *Gyrosan* in  
263 the sagittal plane, allows to observe the movements and deformations of pelvic  
264 organs. Each MRI sequence considered is a subset of 12 frames extracted from  
265 an acquisition during an abdominal pushing down maneuver. The 2D images of  
266 the MRI sequences are segmented in order to produce a set of closed contours  
267 of four organs of interest: bladder, rectum, uterus and vagina. Fig. 7 illustrates  
268 three frames of a MRI sequence, in which the first frame represents the organs  
269 at rest, the last frame shows the condition when the pushing down effort  
270 is at its maximum. From these contours, we can characterize the movement  
271 by calculating the center of mass of each organ, the organ trajectory will  
272 summarize the displacement (Rahim, Bellemare, Pirr o, & Bulot, n.d.). We  
273 can also characterize the deformation by using the shape descriptors (Zhang  
274 & Lu, 2004). Two shapes of organs are compared by computing a distance  
275 between their own descriptor coefficients. The quantification process of the  
276 deformation will not be detailed in this paper, more informations can be found  
277 in (Rahim, Bellemare, Bulot, & Pirr o, 2010). The displacement of contours  
278 between each frame is obtained, which has also been correlated with the nodes  
279 in the contours of our mesh model. By imposing the displacement on each

organ's contour, the contact forces can not be considered, which reveals the inter-penetration between organs. We have also removed the fascia part and the pubis to adapt the new condition. Fig. 8 shows the comparison between the simulation and dynamic MRI images of 12 frames.

## 5 Conclusion and perspectives

In this paper, we focused on the simulations of 2D model of pelvic organs. We have modeled one of the mechanical process occurring within the female pelvic organs in order to improve the understanding of POP. We have simulated the deformation of each organ by imposing the displacement calculated from a MRI sequence of 12 frames during an abdominal pushing down maneuver. The isotropic hyperelastic mechanical behavior is considered for the organ tissues. This is a first step which opens many perspectives and can propose a tool to help physicians to characterize functional impact of these organs. With this tool, it is possible to have a good idea of the contact force, the dimension of contact surface, the deepness of significant stresses, the energy absorbed by the soft tissue, etc.

Biological soft tissues are often assumed to be anisotropic and hyperelastic (Fung, 1993; Holzapfel et al., 2000). Therefore, as a perspective of this work, anisotropic material laws should be adopted in the future. Moreover, a three dimensional model based on MRI data including pelvic organs, pubic bone, as well as ligaments and fascias shall be used in further simulations. To improve the simulation accuracy, more realistic boundary conditions should be considered.



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307 **References**

- 308 Arruda, E. M., & Boyce, M. C. (1993). A three dimensional constitutive model  
309 for the large deformation stretch behavior of rubber elastic materials. *J.*  
310 *Mech. Phys. Solids*, *41*, 389-412.
- 311 Baden, W. F., & Walker, T. (1992). *Surgical repair of vaginal defects*. JB  
312 Lippincott.
- 313 Bay, T., Chen, Z.-W., Raffin, R., Daniel, M., Joli, P., Feng, Z.-Q., & Bellemare,  
314 M. (2012). Geometric modeling of pelvic organs with thickness. In  
315 *Ei2012-electronic imaging*. Burlingame, USA.
- 316 Boubaker, M. B. (2009). *Contribution mécanique à la rduction des marges*  
317 *en radiothérapie de la prostate: modélisation et simulation numérique du*  
318 *mouvement et de la déformation des organes pelviens* (Unpublished doc-  
319 toral dissertation). Institut National Polytechnique de Lorraine, France.
- 320 Boubaker, M. B., Haboussi, M., Ganghoffer, J.-F., & Aletti, P. (2009). Fi-  
321 nite element simulation of interactions between pelvic organs: Predictive  
322 model of the prostate motion in the context of radiotherapy. *J. Biomech.*,  
323 *42*, 1862-1868.
- 324 Boyce, M. C., & Arruda, E. M. (2000). Constitutive models of rubber elas-  
325 ticity: a review. *Rubber Chem. Technol.*, *73*, 504-523.

- 326 Charlton, D. J., & Yang, J. (1994). A review of methods to characterize  
327 rubber elastic behaviour for use in finite element analysis. *Rubber Chem.*  
328 *Technol.*, 67, 481-503.
- 329 Chionh, J. J. L., Wei, B. . P. C., Martin, J. A., & Opdam, H. I. (2006).  
330 Determining normal values for intra-abdominal pressure. *ANZ J. Surg.*,  
331 76, 1106-1109.
- 332 Cobb, W. S., Burns, J. M., Kercher, K. W., Matthews, B. D., Norton, H. J.,  
333 & Heniford, B. T. (2005). Normal intraabdominal pressure in healthy  
334 adults. *J. Surg. Res.*, 129, 231-235.
- 335 DeLancey, J. O. (1992). Anatomic aspects of vaginal eversion after hysterectomy.  
336 *Am. J. Obstet. Gynecol*, 166, 1717-24.
- 337 DeLancey, J. O. (1993). Anatomy and biomechanics of genital prolapse. *Clin.*  
338 *Obstet. Gynecol.*, 36, 897-909.
- 339 Drake, R. L., Vogl, A. W., & Mitchell, A. W. M. (2010). *Gray's anatomie*  
340 *pour les étudiants(2e édition)*. Elsevier-Masson.
- 341 Feng, Z.-Q. (2008). <http://lmee.univ-evry.fr/~feng/FerSystem.html>.
- 342 Feng, Z.-Q., & Feng, Z.-G. (2004). FER/View: An interactive finite element  
343 post-processor. In Z.-H. Yao, M.-W. Yuan, & W.-X. Zhong (Eds.), *Com-*  
344 *putational mechanics*. Beijing, China: Springer.
- 345 Feng, Z.-Q., Peyraut, F., & He, Q.-C. (2006). Finite deformations of Ogden's  
346 materials under impact loading. *Int. J. Non-linear Mech.*, 41, 575-585.
- 347 Feng, Z.-Q., Peyraut, F., & Labeled, N. (2003). Solution of large deformation  
348 contact problems with friction between Blatz-Ko hyperelastic bodies.  
349 *Int. J. Engng. Science*, 41, 2213-2225.
- 350 Feng, Z.-Q., Vallée, C., Fortuné, D., & Peyraut, F. (2006). The 3é hyperelastic  
351 model applied to the modeling of 3D impact problems. *Finite Elem.*  
352 *Anal. Des.*, 43, 51-58.

- 353 Fung, Y. C. (1993). *Biomechanics: Mechanical properties of living tissues*.  
354 Springer.
- 355 Fung, Y. C., Fronek, K., & Patitucci, P. (1979). Pseudoelasticity of arteries  
356 and the choice of its mathematical expression. *Am. J. Physiol.*, *237*,  
357 620-631.
- 358 Gabriel, B., Rubod, C., Brieu, M., Dedet, B., Landsheere, L., Delmas, V., &  
359 Cosson, M. (2011). Vagina, abdominal skin, and aponeurosis: do they  
360 have similar biomechanical properties? *Int. Urogynecol. J.*, *21*, 23-27.
- 361 Gent, A. N. (1996). A new constitutive relation for rubber. *Rubber Chem.*  
362 *Technol.*, *69*, 59-61.
- 363 Hagen, S., Stark, D., & Cattermole, D. (2004). A united kingdom-wide sur-  
364 vey of physiotherapy practice in the treatment of pelvic organ prolapse.  
365 *Physiotherapy*, *90*, 19-26.
- 366 Hall, M. J., DeFrances, C. J., Williams, S. N., Golosinskiy, A., & Schwartzman,  
367 A. (2010). National hospital discharge survey: 2007 summary. *National*  
368 *health statistics reports*, *29*, 1-24.
- 369 Harth-Smith, J. L. (1996). Elasticity parameters for finite deformations of  
370 rubber-like materials. *Math. Phys.*, *17*, 608-625.
- 371 Hendrix, S. L., Clark, A., Nygaard, I., Aragaki, A., Barnabei, V., & McTier-  
372 nan, A. (2002). Pelvic organ prolapse in the women's health initiative:  
373 gravity and gravidity. *Am. J. Obstet. Gynecol.*, *186*, 1160-1166.
- 374 Holzapfel, G. A. (2006). *Mechanics of biological tissue*. Springer.
- 375 Holzapfel, G. A., Gasser, T. C., & Ogden, R. (2000). A new constitutive  
376 framework for arterial wall mechanics and a comparative study of ma-  
377 terial models. *J. Elasticity*, *61*, 1-48.
- 378 Jackson, S., & Smith, P. (1997). Diagnosing and managing genitourinary  
379 prolapse. *Br. Med. J.*, *314*, 875-880.

- 380 Jean-Charles, C., Rubod, C., Brieu, M., Boukerrou, M., Fasel, J., & Cosson,  
381 M. (2010). Biomechanical properties of prolapsed or non-prolapsed  
382 vaginal tissue: impact on genital prolapse surgery. *Int. Urogynecol. J.*,  
383 *21*, 1535-1538.
- 384 Lambert-Diani, J., & Rey, C. (1998). New phenomenological laws for rubber  
385 and thermoplastic elastomers. *Eur. J. Mech.*, *1(18-6)*, 1027-1043.
- 386 Luft, J. (2006). Organ prolapse: current state of knowledge about this common  
387 condition. *J. Nurse Pract.*, *2(3)*, 170-177.
- 388 Messelink, B., Benson, T., Berghmans, B., Bø, K., Corcos, J., Fowler, C., ...  
389 Von Kerrebroeck, P. (2005). Standardization of terminology of pelvic  
390 floor muscle function and dysfunction: report from the pelvic floor clinical  
391 assessment group of the international continence society. *Neurourol.*  
392 *Urodynam.*, *24*, 374-380.
- 393 Miehe, C., Göktepe, S., & Lulei, F. (2004). A micro-macro approach to  
394 rubber-like materials part i: The non-affine micro-sphere model of rubber  
395 elasticity. *J. Mech. Phys. Solids*, *52*, 2617-2660.
- 396 Mooney, M. (1940). A theory of large elastic deformation. *J. Appl. Phys.*,  
397 *11(9)*, 582-592.
- 398 Noakes, K. F., Pullan, A. J., Bissett, I. P., & Cheng, L. K. (2008). Subject  
399 specific finite elasticity simulations of the pelvic floor. *J. Biomech.*, *41*,  
400 3060-3065.
- 401 Olsen, A. L., Smith, V. J., Bergstrom, J. O., Colling, J. C., & Clark, A. (1997).  
402 Epidemiology of surgically managed pelvic organ prolapse and urinary  
403 incontinence. *Obstet. Gynecol.*, *89*, 501-506.
- 404 Pena, E., del Palomar, A. P., Calvo, B., Martínez, M. A., & Doblar, M. (2007).  
405 Computational modelling of diarthrodial joints. physiological, patholog-  
406 ical and post-surgery simulations. *Arch. Comput. Meth. Engrg*, *14(1)*,

407 1-54.

408 Pena, E., Martínez, M. A., Calvo, B., & Doblar, M. (2006). On the numerical  
409 treatment of initial strains in soft biological tissues. *Int. J. Numer. Meth.*  
410 *Engrg*, 68, 836-860.

411 Pirró, N., Bellemare, M. E., Rahim, M., Durieux, O., Sielezneff, I., Sastre, B.,  
412 & Champsaur, P. (2009). Résultats préliminaires et perspectives de la  
413 modélisation dynamique pelvienne patient-spécifique. *Pelv. Perineol.*,  
414 4, 15-21.

415 Rahim, M., Bellemare, M. E., Bulot, R., & Pirró, N. (2010). Pelvic organs dy-  
416 namic feature analysis for MRI sequence discrimination. In *International*  
417 *conference on pattern recognition*. Istanbul.

418 Rahim, M., Bellemare, M. E., Pirró, N., & Bulot, R. (n.d.). A shape descriptors  
419 comparison for organs deformation characterization in MRI sequences..

420 Rivlin, R. S. (1948). Large elastic deformations of isotropic materials. IV.  
421 Further developments of the general theory. *Phil. Trans. R. Soc. Lond.*  
422 *A*, 241(835), 379-397.

423 Rubod, C., Boukerrou, M., Brieu, M., Dubois, P., & Cosson, M. (2006a).  
424 Biomechanical properties of vaginal tissue. part 1: New experimental  
425 protocol. *J. Urol.*, 178, 320-325.

426 Rubod, C., Boukerrou, M., Brieu, M., Dubois, P., & Cosson, M. (2006b).  
427 Biomechanical properties of vaginal tissue. Part 1: New experimental  
428 protocol. *J. Urol.*, 178, 320-325.

429 Schröder, J., Neff, P., & Balzani, D. (2005). A variational approach for  
430 materially stable anisotropic hyperelasticity. *Int. J. Solids. Struct.*, 42,  
431 4352-4371.

432 Spencer, A. J. M. (1954). Theory of invariants. *Continuum Physics, Academic*  
433 *Press, New York*, 239-253.

- 434 Thakar, R., & Stanton, S. (2002). Management of genital prolapse. *Br. Med.*  
435 *J.*, *324(7348)*, 1256-1262.
- 436 Venugopala Rao, G., Rubod, C., Brieu, M., Bhatnagar, N., & Cosson, M.  
437 (2010). Experiments and finite element modelling for the study of pro-  
438 lapse in the pelvic floor system. *Comput. Methods Biomech. Biomed.*  
439 *Eng.*, *13*, 349-357.
- 440 Weiss, J. A., Maker, B. N., & Govindjee, S. (1996). Finite element implemen-  
441 tation of incompressible, transversely isotropic hyperelasticity. *Comput.*  
442 *Meth. Appl. Mech. Engng*, *135*, 107-128.
- 443 Zhang, D., & Lu, G. (2004). Review of shape representation and description  
444 techniques. *Pattern Recogn.*, *37*, 1-19.

Table 1

Material parameters of organs according to Yeoh strain energy density

Material coefficients	Vagina(uterus)	Rectum	Bladder
$C_{10}$ (MPa)	0.175	0.088	0.071
$C_{20}$ (MPa)	8.648	3.092	0.202
$C_{30}$ (MPa)	8.632	2.871	0.048
$K$ (MPa)	16.0	11.0	9.0

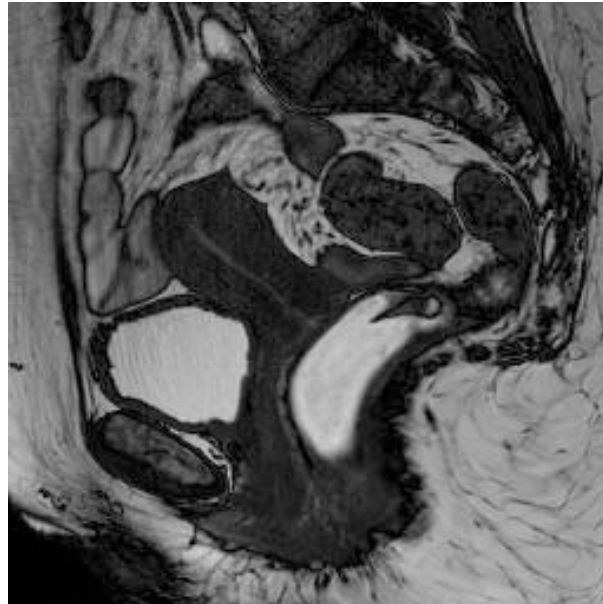


Fig. 1. MRI image for the development of the geometric model

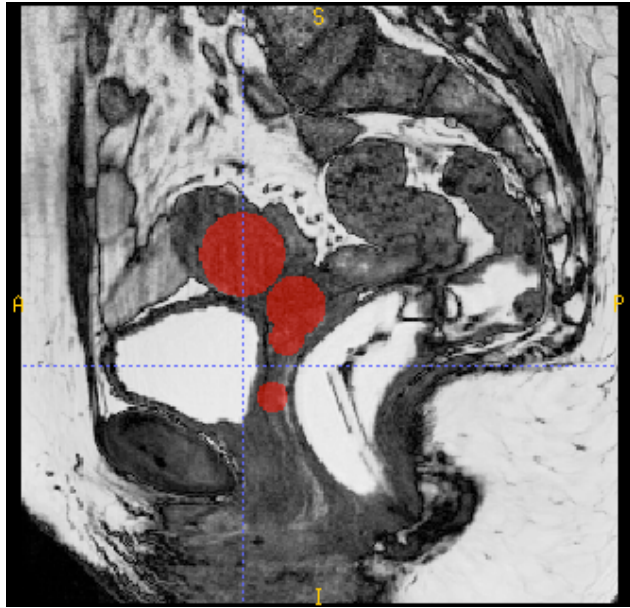


Fig. 2. Volumetric germs placed inside of the uterus and vagina

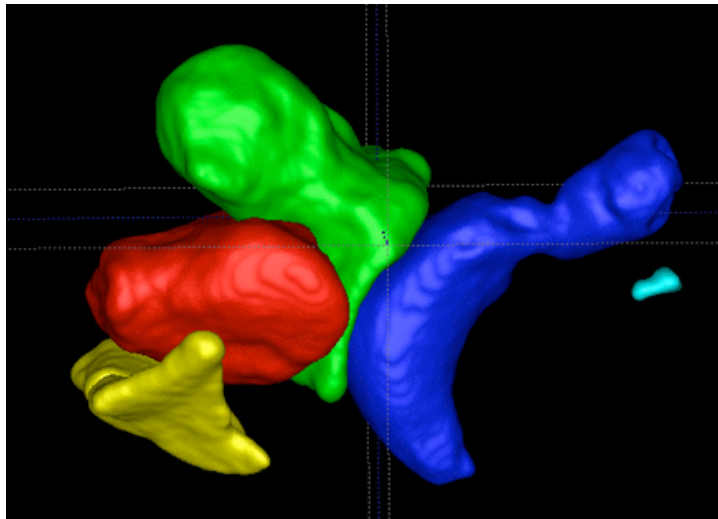


Fig. 3. Complete geometric model





Fig. 4. Schematic location of pelvic organs in the pelvic cavity

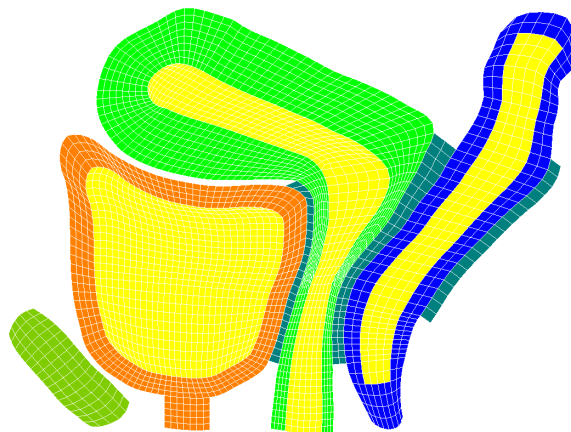


Fig. 5. Initial configuration and meshes

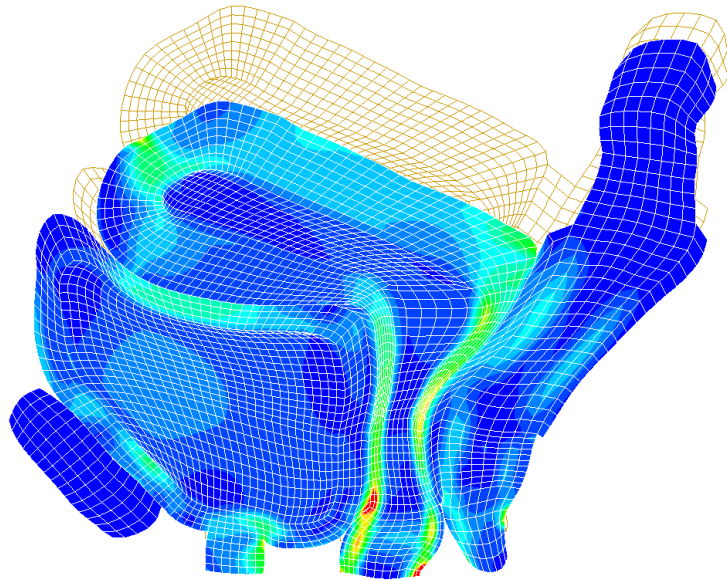


Fig. 6. Initial configuration and deformed shape with von Mises stress

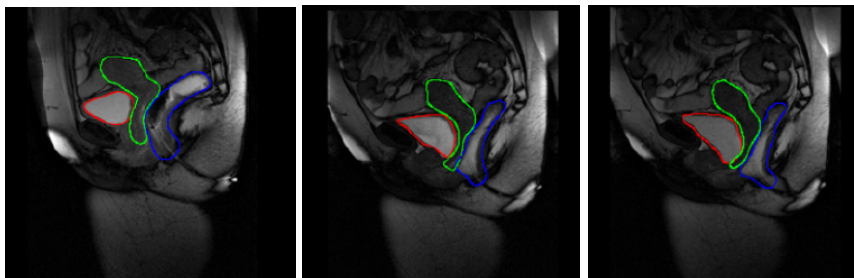


Fig. 7. Segmented MRI sequence

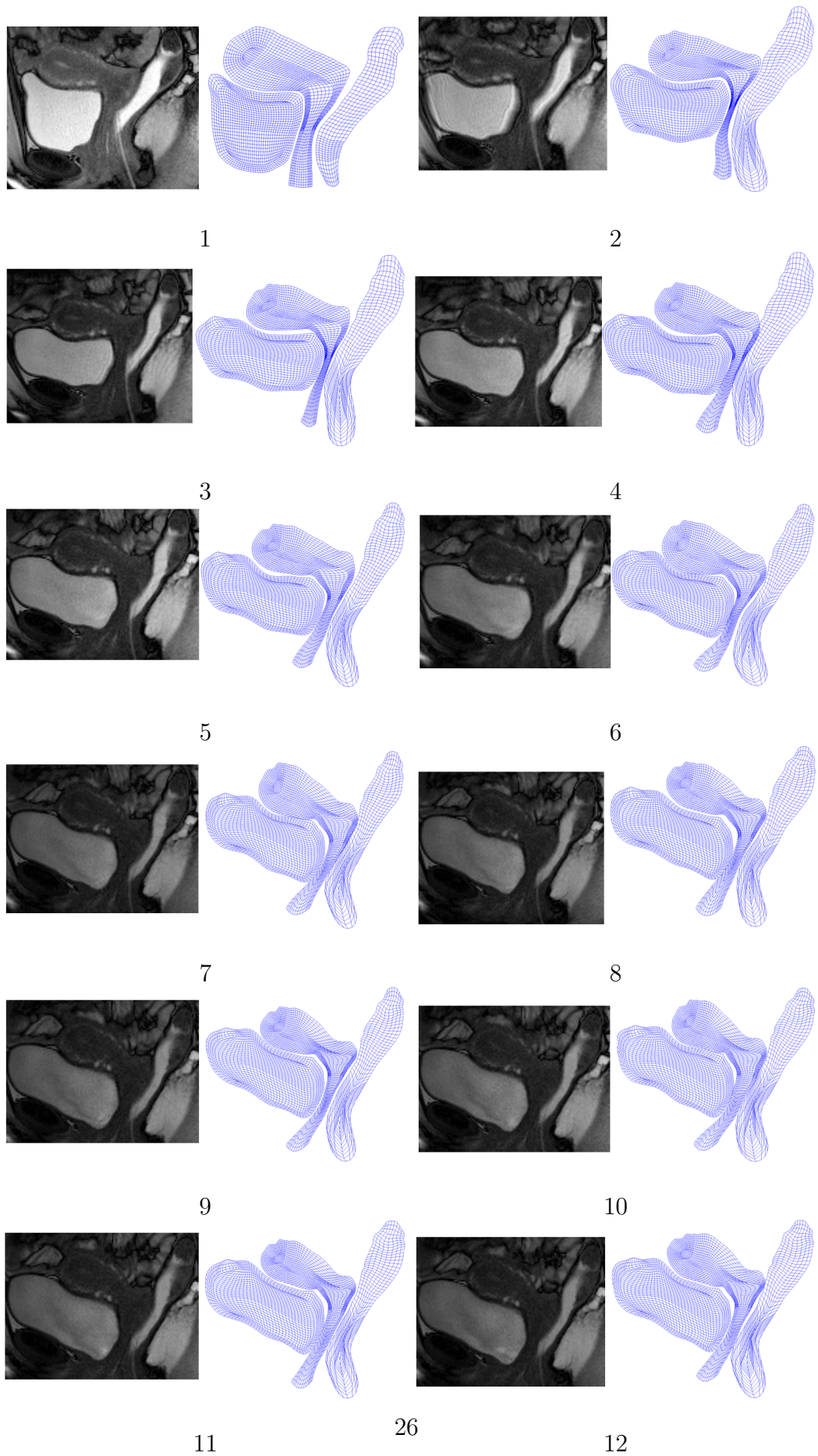


Fig. 8. Comparison of the deformation between numerical results and MRI images