



HAL
open science

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

Zhuowei Chen, Pierre Joli, Zhi-Qiang Feng, Mehdi Rahim, Nicolas Pirr6,,
Marc-Emmanuel Bellemare

► **To cite this version:**

Zhuowei Chen, Pierre Joli, Zhi-Qiang Feng, Mehdi Rahim, Nicolas Pirr6,, et al.. Female patient-specific finite element modeling of pelvic organ prolapse (POP). 2024. hal-01176093

HAL Id: hal-01176093

<https://hal.science/hal-01176093>

Preprint submitted on 1 Feb 2024

HAL is a multi-disciplinary open access archive for the deposit and dissemination of scientific research documents, whether they are published or not. The documents may come from teaching and research institutions in France or abroad, or from public or private research centers.

L'archive ouverte pluridisciplinaire **HAL**, est destinée au dépôt et à la diffusion de documents scientifiques de niveau recherche, publiés ou non, émanant des établissements d'enseignement et de recherche français ou étrangers, des laboratoires publics ou privés.

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

Z.-W. Chen ^a, P. Joli ^a, Z.-Q. Feng ^{a,b,*}, M. Rahim ^c, N. Pirró ^d,

M. E. Bellemare ^c

^a*Laboratoire de Mécanique et d'Energétique d'Évry, Université d'Évry-Val
d'Essonne, Évry, France*

^b*School of Mechanics and Engineering, Southwest Jiaotong University, Chengdu,
China*

^c*Laboratoire des Sciences de l'Information et des Systèmes, Marseille, France*

^d*Service de Chirurgie Digestive, Hôpital la Timone, Marseille, France*

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

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

* Corresponding author. Tel.: +33 1 69 47 75 01; Fax: +33 1 69 47 75 99.

E-mail address: feng@ufrst.univ-evry.fr (Z.-Q. Feng)

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,
98 256×256 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, 256×256 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 , & 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 , 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.

303 **Acknowledgement**

304 This work is supported by the French ANR Grant (No. ANR-09-SYSC-008)
305 and the National Natural Science Foundation of China (No. 11372260), which
306 are greatly appreciated.

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

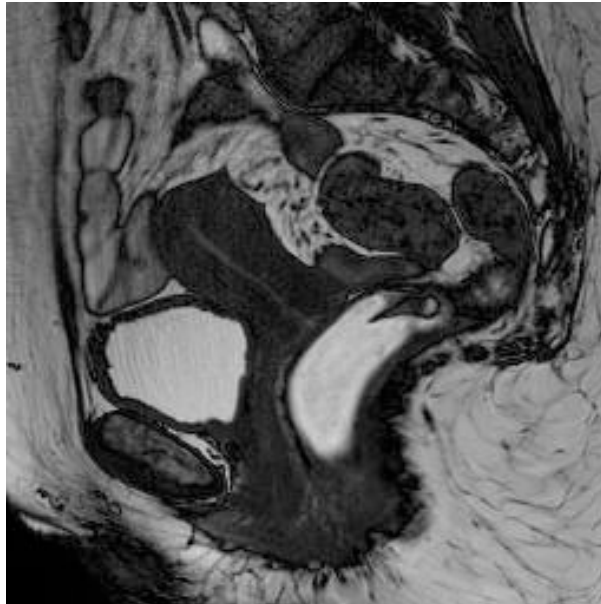


Fig. 1. MRI image for the development of the geometric model

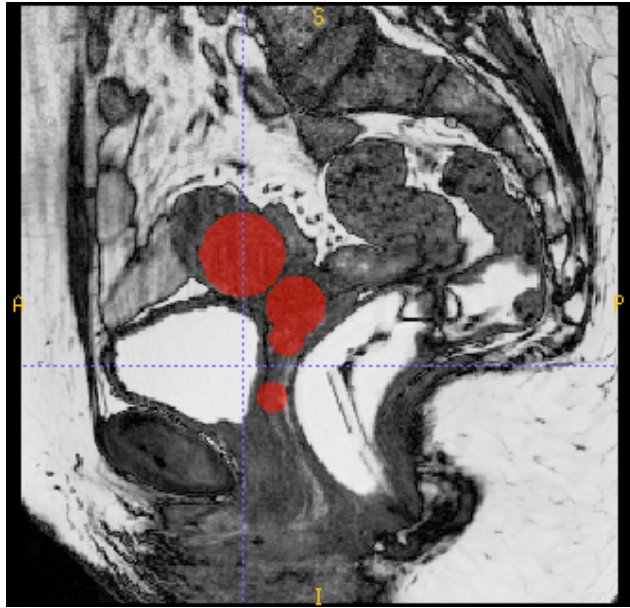


Fig. 2. Volumetric germs placed inside of the uterus and vagina

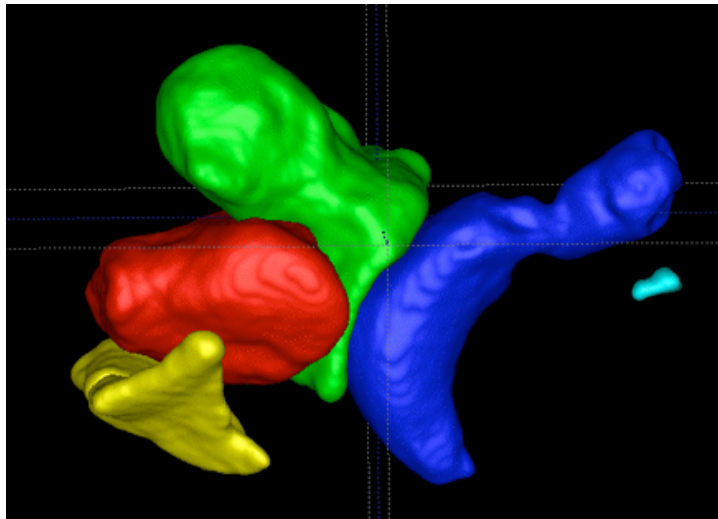


Fig. 3. Complete geometric model



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

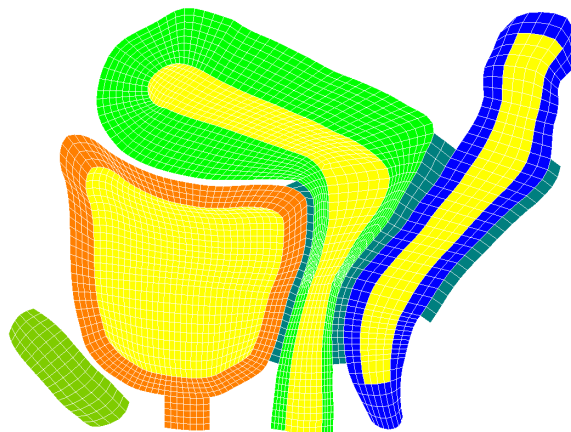


Fig. 5. Initial configuration and meshes

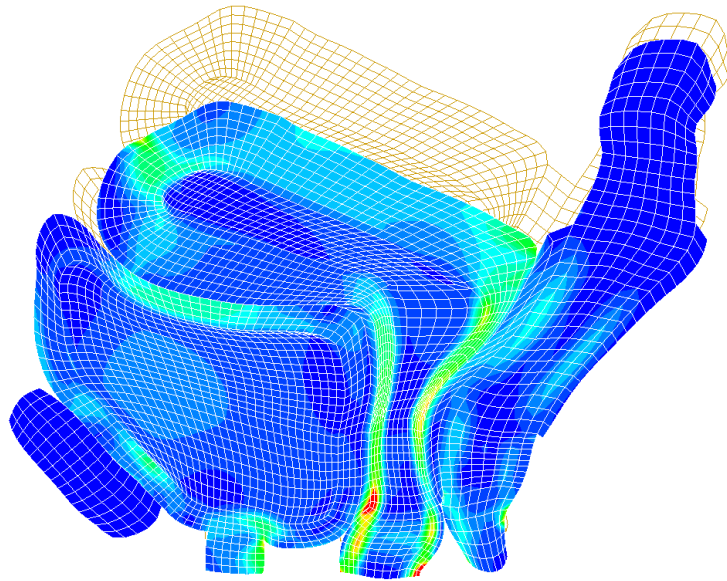


Fig. 6. Initial configuration and deformed shape with von Mises stress

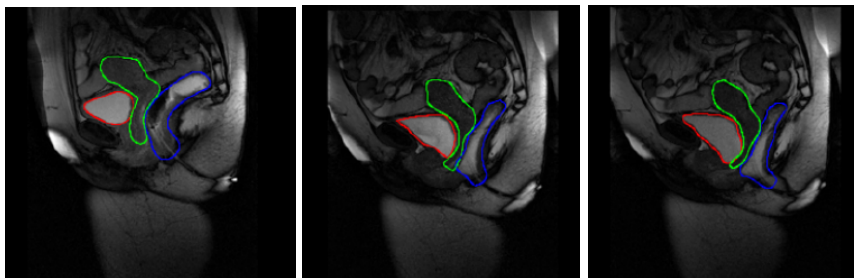


Fig. 7. Segmented MRI sequence

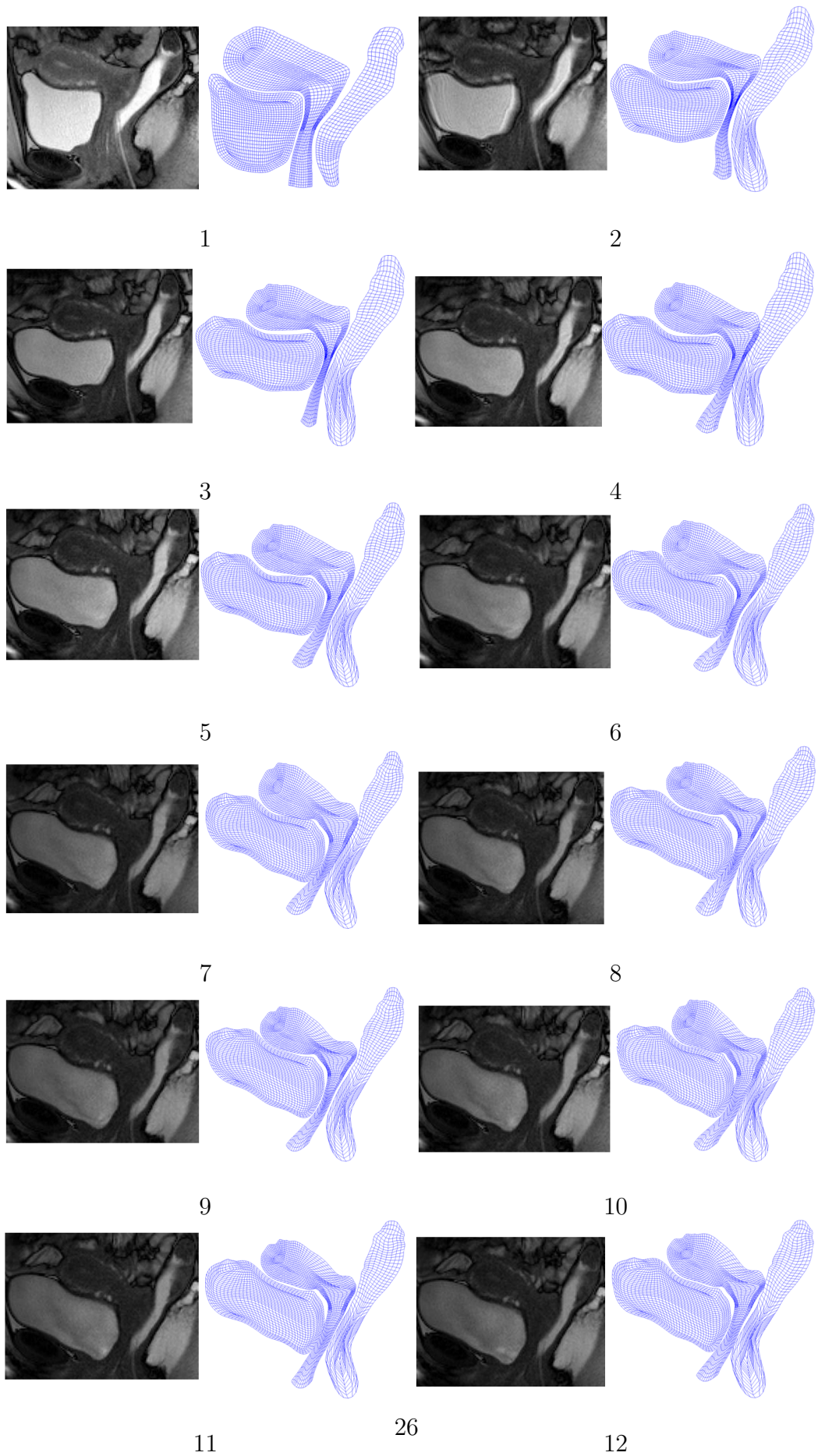


Fig. 8. Comparison of the deformation between numerical results and MRI images