

# Prostheses size dependency of the mechanical response of the herniated human abdomen

R. Simón-Allué<sup>1,2</sup> · B. Hernández-Gascón<sup>1,2</sup> · L. Lèoty<sup>3</sup> · J. M. Bellón<sup>2,4</sup> · E. Peña<sup>1,2</sup> · B. Calvo<sup>1,2</sup>

Received: 17 March 2016 / Accepted: 29 July 2016 / Published online: 22 August 2016  
© Springer-Verlag France 2016

## Abstract

**Background** Hernia repairs still exhibit clinical complications, i.e. recurrence, discomfort and pain and mesh features are thought to be highly influent. The aim of this study is to evaluate the impact of the defect size and mesh type in an herniated abdominal wall using numerical models.

**Methods** To do so, we have started from a FE model based on a real human abdomen geometry obtained by MRI, where we have provoked an incisional hernia of three different sizes. The surgical procedure was simulated by covering the hernia with a prostheses, and three surgical meshes with distinct mechanical properties were used for the hernia repair: an isotropic heavy-weight mesh (*Surgipro*<sup>®</sup>), a slightly anisotropic light-weight mesh (*Optilene*<sup>®</sup>) and a highly anisotropic medium-weight mesh (*Infini*<sup>®</sup>). The mechanical response of the wall to a high intraabdominal pressure (corresponding to a coughing motion) was analyzed here.

**Results** Our findings suggest that the anisotropy of the mesh becomes more relevant with the increase of the defect size. Additionally, according to our results *Optilene*<sup>®</sup> showed the closest deformation to the natural distensibility

of the abdomen while *Infini*<sup>®</sup> should be carefully used due to its excessive compliance.

**Keywords** Hernia · Repair · Polypropylene mesh · Failure · Finite element analysis

## Introduction

Nowadays, clinical evidence recommends the use of prosthetic materials for hernia repair, being this a prime defect (primary hernia) or derived from a previous laparotomy (incisional hernia). Last years, the evolution of prosthetic materials has raised meshes with different spatial arrangement of filaments and specially pore size [1, 2]. Consequently, porosity has led to classify meshes in three distinct groups: high (HW, above 80 g/m<sup>2</sup>), medium (MW, between 80 and 50 g/m<sup>2</sup>) or low density (LW, below 50 g/m<sup>2</sup>). Pore size is linked to the prosthetic weight (g/m<sup>2</sup>): small pore size is related to the high density while large size is associated to the low density meshes [3].

After repairing an abdominal hernia defect with a surgical mesh, it is necessary to ensure that the implant can bear the abdominal mechanical loads the post-implant is going to suffer. To do so, the great variability of the intraabdominal pressure (IAP) has to be considered [4]. Moreover, once the mesh is placed into the tissue the whole has to work in the most physiological way, which is specially hard to determine since this mechanical demand varies depending on the individual biotype. Regarding this aspect, obesity has revealed to be a very determining factor [4].

In general, high density meshes are associated to a high tensile strength, but also a profound tissue reaction and dense scarring due to the large mass of material [3]. Low density, on the contrary, attempt to produce less

✉ R. Simón-Allué  
rsimon@unizar.es

<sup>1</sup> Applied Mechanics and Bioengineering, Aragón Institute of Engineering Research, University of Zaragoza, Saragossa, Spain

<sup>2</sup> CIBER de Bioingeniería, Biomateriales y Nanomedicina (CIBER-BBN), Madrid, Spain

<sup>3</sup> Ecole Centrale de Centrale Nantes, Nantes, France

<sup>4</sup> Department of Surgery, Medical and Social Sciences, Faculty of Medicine and Health Sciences, Universidad de Alcalá, Madrid, Spain

pronounced foreign body reaction and are considered to be more compliant. During the last decade and owing to the higher use of LW meshes, some cases of mesh breakage have been reported after hernia repairs. Although first description arose from a patient repaired with a HW [5], successive cases are associated to the use of low density meshes [6, 7] or partially absorbable [8, 9]. Also in our group, previous mechanical findings revealed the risk of breakage of the medium weight ( $70 \text{ g/m}^2$ ) *Infini*<sup>®</sup> mesh (INF) when a high tensile strength is required [10].

Numerous prior studies have focused their work on examining the response when using prosthetic materials [11–13]. However, it is a really complex task to experimentally address the compliance and stress generated to the tissues and implant. This inconvenient could be solved through the use of numerical models. Several authors have made use of finite element models applied to idealized geometries to study the mechanical response of human tissues, some of them focused on the inguinal region [14, 15]. Guérin et al. [16] evaluated the impact of the defect size, the mesh overlap and the fixation depth on ventral hernia repairs using a simplified numerical models and Hernández-Gascón et al. [10] developed a FE model of an herniated human abdominal wall to simulate the response of different prosthesis and the tissue when a high intra abdominal pressure is generated. However, only one mesh size was analyzed there. In this work, we employed this model to compare the mechanical response of different meshes (*Surgipro*<sup>®</sup> (SUR), an isotropic heavy-weight mesh; *Optilene*<sup>®</sup> (OPT), a slightly anisotropic light-weight mesh and *Infini*<sup>®</sup> (INF), a highly anisotropic medium-weight mesh) after their implant over different size of incisional hernia (small:  $25 \times 50 \text{ mm}$ , medium:  $60 \times 120 \text{ mm}$  and large:  $120 \times 200 \text{ mm}$ ). The surgical

procedure of hernia repair is simulated for three different prostheses to analyze immediate stress states produced in the prostheses and surroundings under a high pressure value in the most unfavorable conditions, prior to the tissue regeneration when the mesh is working alone.

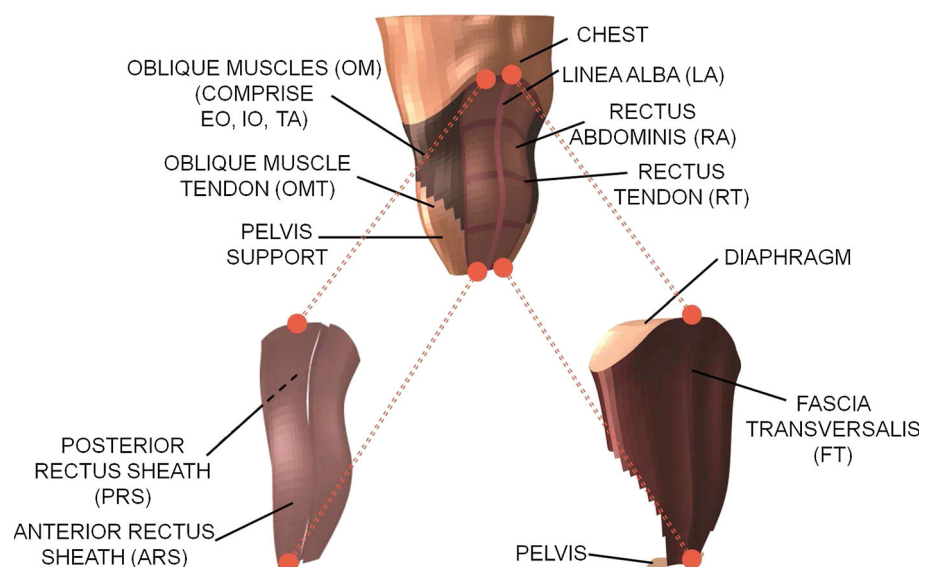
## Methods

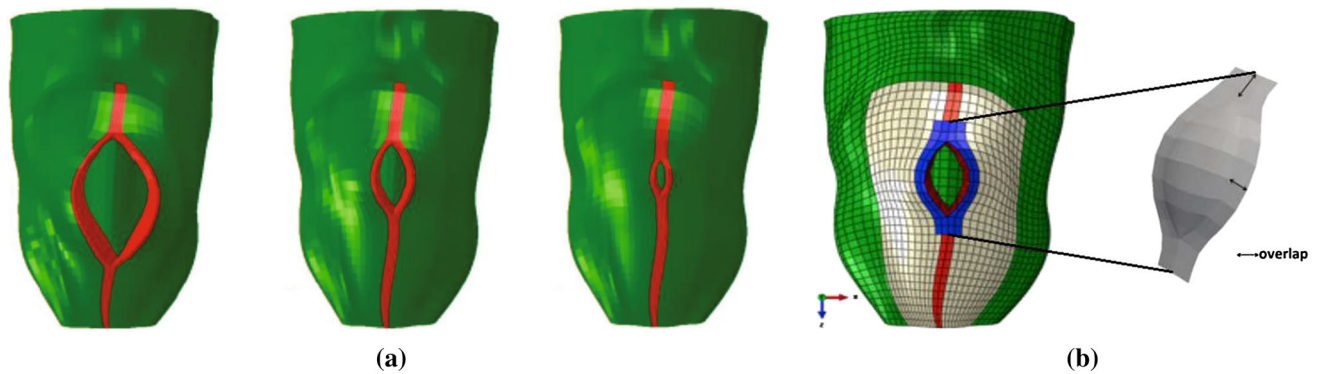
A FE model of an herniated abdominal wall was reconstructed based on a previous FE model of the human abdomen of a healthy 38-year-old man [17]. In this model, different anatomical structures were considered according to their contribution to stiffness of the whole. In this sense, muscles and aponeuroses were included in the model while fat and skin were excluded for not being considered as determinant in terms of stiffness [18].

The structures distinguished by the FE model were: linea alba (LA), rectus tendon (RT), rectus abdominis muscle (RAM), oblique muscles (comprising the external and internal obliques and the transversus abdominis), oblique muscle tendon (OMT), fascia transversalis (FT), anterior and posterior rectus sheath (ARS and PRS, respectively), diaphragm, chest, back and pelvis (see Fig. 1).

The FE model and the defect were created in supine position since DICOM images were taken with the subject lying down [17]. Three different sizes of incisional hernia were considered in this work to assess the prostheses size dependency of the mechanical response of the herniated abdomen: large ( $120 \times 200 \text{ mm}$ ), a medium ( $60 \times 100 \text{ mm}$ ) and small ( $25 \times 50 \text{ mm}$ ) (see Fig. 2a) [20]. First dimension refers to the maximum width found in the midline of the defect, while second concerns the maximum length measured between the nodes at the top and at the

**Fig. 1** Finite element model of the healthy human abdomen where the different anatomical structures are identified. Since different structures are superimposed anatomically, they are shown separately and the *dotted lines* indicate the geometrical correspondence [19]





**Fig. 2** **a** FE model of the herniated human abdomen for the three size considered: large ( $120 \times 200$  mm); medium ( $60 \times 100$  mm); small ( $25 \times 50$  mm). **b** Overlap (painted in blue in the figure) between prosthesis and tissue

bottom of the defect. The linea alba in Fig. 2a is represented in red. The process of generating the defect into the FE model of the healthy abdomen is fully detailed in Hernández-Gascón et al. [12].

In the model we simulated onlay mesh technique, where the surgical mesh is placed covering the hernia defect on the external surface of the aponeurosis. The prosthesis was modeled with the software *ANSYS ICEM CFD*<sup>®</sup>, using membrane elements of 1 mm of thickness [12]. This value has to be assigned because of the stress stretch curves, which are determined by depth unit. The use of the onlay technique implies an overlap between the prosthesis and abdominal wall (see Fig. 2b). This overlap was recreated by placing some elements on the surrounding tissue, in particular 3 elements on the lateral sides and 2 elements on the upper and lower sides of the overlap. To suture the surgical mesh, a double continue suture was modeled here by matching nodes corresponding to the prosthesis and abdomen.

To define the material properties through constitutive modeling, the purely anisotropic [21] passive response of the abdominal muscle was modeled using an anisotropic exponential-linear strain energy function that considers the direction of the muscular and collagen fibers. The anisotropic behavior associated with the passive response is modeled by introducing a preferential direction of anisotropy (PDA), defined by the angle  $\alpha$  considered from the crano-caudal direction. Depending on the abdomen component, this angle was obtained by a fitting procedure or following the literature or assuming that collagen fibers have the same disposition as the muscle fibers, as in the case of fascias. Diaphragm and pelvis were modeled with a Neo-Hookean model. A fully description of the anisotropic model as well as the material properties were taken from prior works [12, 22, 23].

Due to the anisotropy of the meshes OPT and INF, the mechanical response of the abdominal model may vary depending on the orientation of these meshes with these to

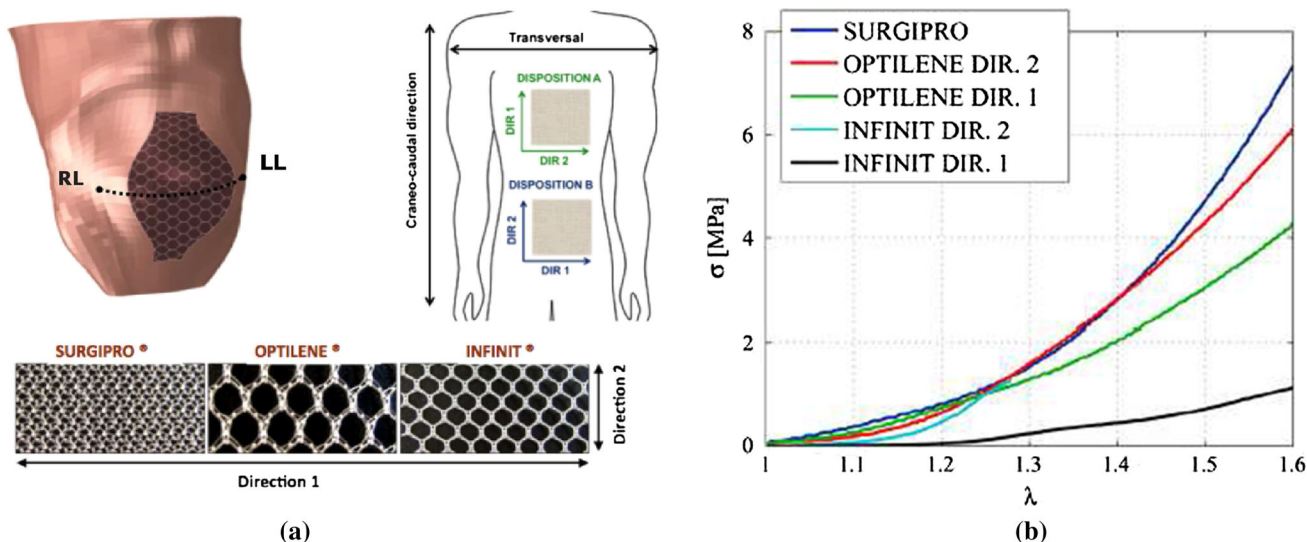
the abdomen. Thus, two principal directions were considered in this meshes to examine the effects of its anisotropy, direction 1 and 2 (see Fig. 3a), defined in prior works by Hernández-Gascón et al. [12]. For anisotropic meshes, direction 1 agreed with the most compliant direction while direction 2 remained as the stiffest. Considering this, we defined two mesh dispositions on the abdomen depending on the mesh orientation. In Disposition A, direction 1 was aligned with the crano-caudal direction while in Disposition B, direction 1 lay parallel to the transverse direction (see Fig. 3a). As a result, the stiffest axis of the prostheses laid perpendicular to the linea alba for orientation A and parallel for orientation B.

To define the material properties of the meshes two different material models were considered: Demiray's model in the case of the SUR mesh (isotropic response) and Demiray–Holzapfel's model for the OPT and INF meshes (anisotropic response). Material parameters of each mesh were taken from prior works [12]. Fibers were considered along direction 2 of the meshes since it remained as the stiffest direction.

An additional line was considered to analyze results of this study, located through the abdominal wall in the transversal direction from right lateral (RL) to left lateral (LL) (see Fig. 3a).

To determinate the boundary conditions of the model, the physiology of the trunk was considered. When the abdomen is subjected to a high intraabdominal pressure (IAP), the movement of the dorsal part is restricted by the ribs. This constraint was including in the model by fixing the nodes in the dorsal abdomen and pelvis.

The whole process was performed in two load steps. During the first one, the geometry was modeled to change the position from supine to standing. Both body mass (gravity =  $4.64 \cdot 10^{-6} \frac{\text{kg}}{\text{mm}^3}$ ), and the weight of the viscera and muscles were considered for this step, as well as an IAP of 20 mmHg for the standing position [4]. Since IAP has been considered an influencing factor in cases related



**Fig. 3** **a** Orientations defined to place the mesh with respect to the abdomen: *Orientation A* The most compliant direction of the mesh, direction 1, is aligned with the craneo-caudal axis. *Orientation B* Direction 1 is aligned with the transverse direction. Path results along line RL–LL. **b** Experimental data from uniaxial mechanical tests conducted on the prostheses examined obtained from the literature [10]

to trauma or abdominal diseases, we decided to simulate this effect. Thus, during the second load step the IAP was increased to IAP = 172.5 mmHg, corresponding to the physiological IAP of a jumping motion.

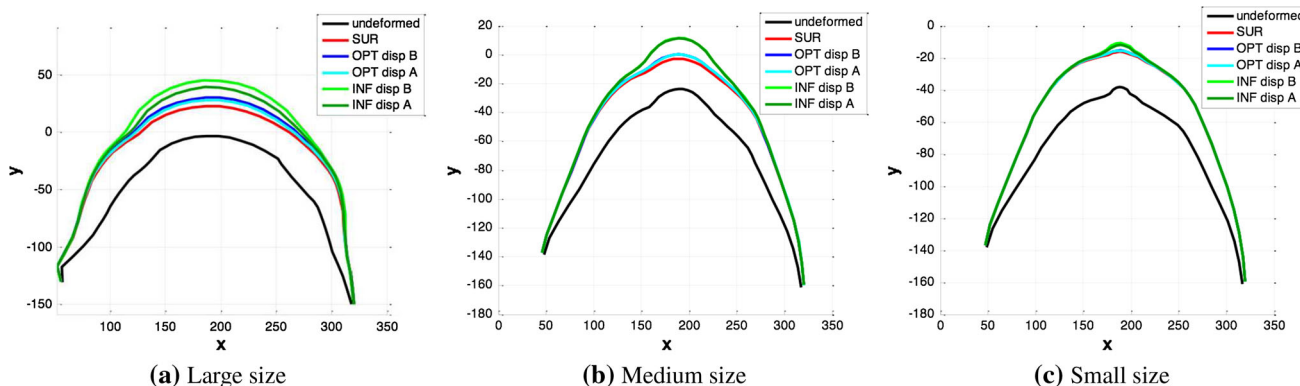
The software used to conduct the FE simulation was ABAQUS v.11<sup>®</sup> in an implicit formulation.

**Results**

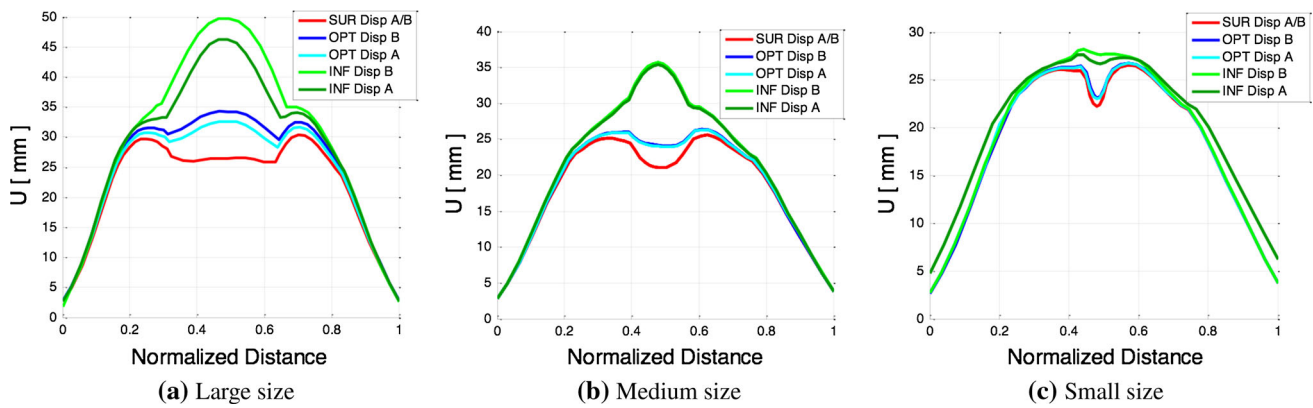
Figure 4 shows the deformed shape for the three defect sizes and the three prostheses simulated, along the line RL–LL defined in Fig. 3a, in the most external anatomical structures. X and Y axis correspond to the coordinates (in mm) for the path RL–LL, used in the FE model. The undeformed shape, which represents the repaired abdomen before applying an

internal pressure, was added to see the initial curvature of the abdomen and to analyze how the curvature changes depending on the defect size, the implanted mesh and the mesh disposition (A or B, see Fig. 3a). Note that the undeformed shape the defect area shows a slightly initial deformation, this effect is called bulging [24]. A jumping movement considerably increase the curvature of the abdomen since it is an extreme intraabdominal pressure [4]. According to Fig. 4, whatever the size defect, the results did not show significant differences. If we analyze the different prostheses, we can find a general tendency: SUR mesh lead to the smallest deformations; OPT mesh presented slightly higher deformations and finally INF mesh showed greater deformations.

Besides, concerning the anisotropic meshes OPT and INF, the transversal curvature of the abdomen was higher with disposition B. As previously introduced and according



**Fig. 4** Profile curves of the deformed shape obtained along the path RL–LL. X and Y axis correspond to the coordinates (in mm) used in the FE model



**Fig. 5** Displacements (MD) along the path RL–LL. The abscissa shows the normalized distance of line RL–LL:  $x = 0$  correspond to point RL and  $x = 1$  to point LL in Fig. 3

**Table 1** Experimental breaking results and maximal stretch and principal stress obtained in the simulation for the different prostheses and sizes (breaking results obtained from literature [13] for uniaxial tests)

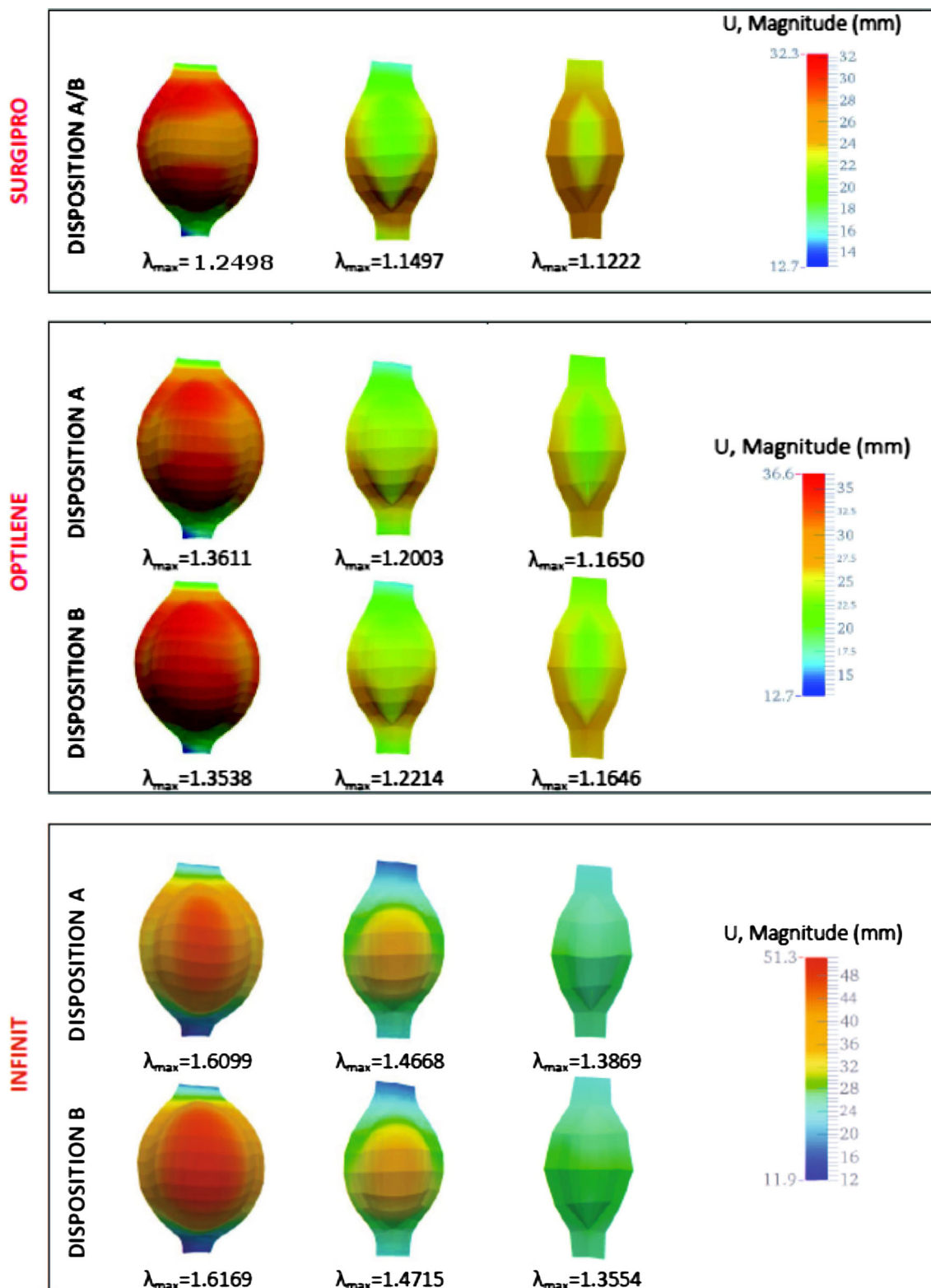
Mesh	Experimental results [13]		Numerical results						
	Breaking stretch (–)	Breaking stress (Mpa)	Stretch [–]			Stress (MPa)			
			Small	Medium	Large	Small	Medium	Large	
Surgipro	–	1.96	19.20	1.122	1.150	1.250	1.012	1.380	2.835
Optilene	Orientation A	1.82	6.359	1.165	1.200	1.361	0.965	1.519	2.940
	Orientation B	1.81	9.560	1.164	1.221	1.354	0.875	1.295	2.738
Infinite	Orientation A	1.76	1.802	1.387	1.467	1.610	1.018	2.272	3.972
	Orientation B	1.31	1.204	1.355	1.471	1.617	0.662	1.329	2.619

to Fig. 3, disposition B assumes that the stiffest direction is aligned with the craneo-caudal direction of the abdomen. Thus, our results suggest that the stiffest direction of the mesh should be aligned with the transversal direction of the abdomen since the compliance between tissue and mesh is more appropriate.

Figure 5 shows the maximal displacements in mm (MD) found along the line RL–LL when large (a), medium (b) and small (c) defects are simulated. The abscissa shows the normalized distance of line RL–LL, assuming  $x = 0$  to point RL and  $x = 1$  to point LL. Among the three prostheses, SUR mesh seemed to lead in a maximal restriction of the natural movement of the abdomen. Displacements in SUR and OPT meshes along the small and the medium prostheses were slightly smaller than healthy abdominal wall. It means that these surgical meshes restrict movement in the repaired defect because of their high stiffness. For large defects, SUR mesh also restricted the movement in the defect area whereas OPT mesh seems to be closer to the natural distensibility of the abdomen. Concerning INF mesh, displacements recorded largely exceed the natural displacements of the abdomen in all cases, which points out that this mesh is too compliant and may not be capable of maintaining the pull out forces generated by viscera in the abdomen.

On the borders of the defect, the effect of the overlap zone can be noticed for medium and large sizes. In cases of INF mesh and due to the great compliance of this mesh, the overlap provoked a point of stiffness on the sides of the defect which could lead to discomfort in the operated patient. Something similar happened for OPT meshes with the large defect but to a lesser extent. Conversely, for SUR meshes and OPT meshes with the medium size, this effect disappeared since the mesh seemed to provoke a stiffer response than the overlap area. No overlap effect was visible for any mesh in case of small size defect.

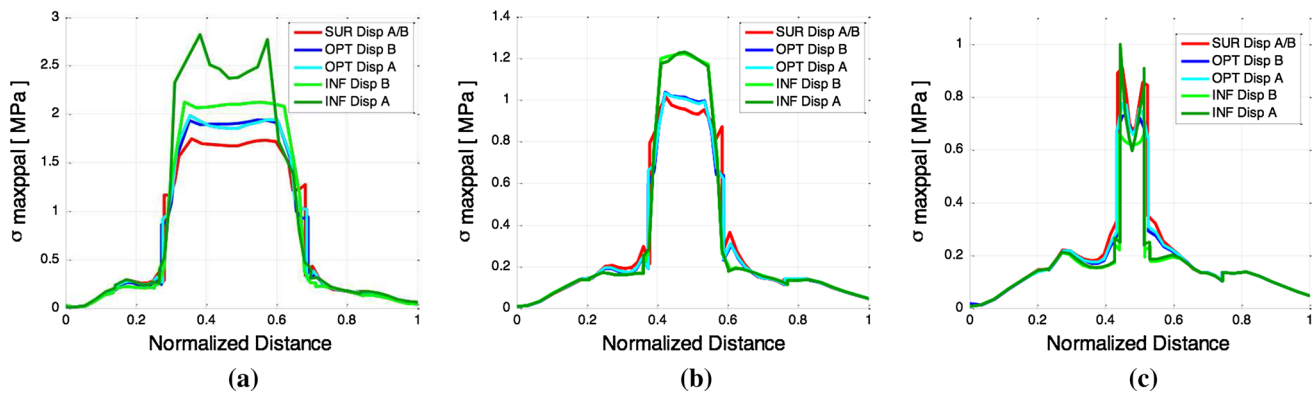
From other point of view, it is also interesting to analyze the mechanical response of the different prostheses depending on the disposition to determine the influence of the placement on the abdomen when the mesh is anisotropic. Main differences were found for the large defect whereas the medium and small defects addressed the same results for both dispositions. This is due to the fact that, in large defect, the abdomen and prosthesis reached higher stretches. For this size, disposition B leads to a higher distension of the abdomen. Maximum stretch values recorded for the different prostheses are presented in Table 1.



**Fig. 6** Displacements (mm) produced in the prostheses after a jumping motion just after surgery

Results in displacement maps shown in Fig. 6 confirm the previous observations. In this figure, displacements produced in the whole prostheses are shown together with

values of maximum stretches. Reinforcing prior results, maximum displacements were lower for SUR mesh because of its great stiffness while INF mesh showed the



**Fig. 7** MPS along the line RL–LL in the model of the herniated abdomen just after surgery (see Fig. 3). The abscissa shows the normalized distance of the line RL–LL.  $x = 0$  and  $x = 1$  correspond to points RL and LL, respectively

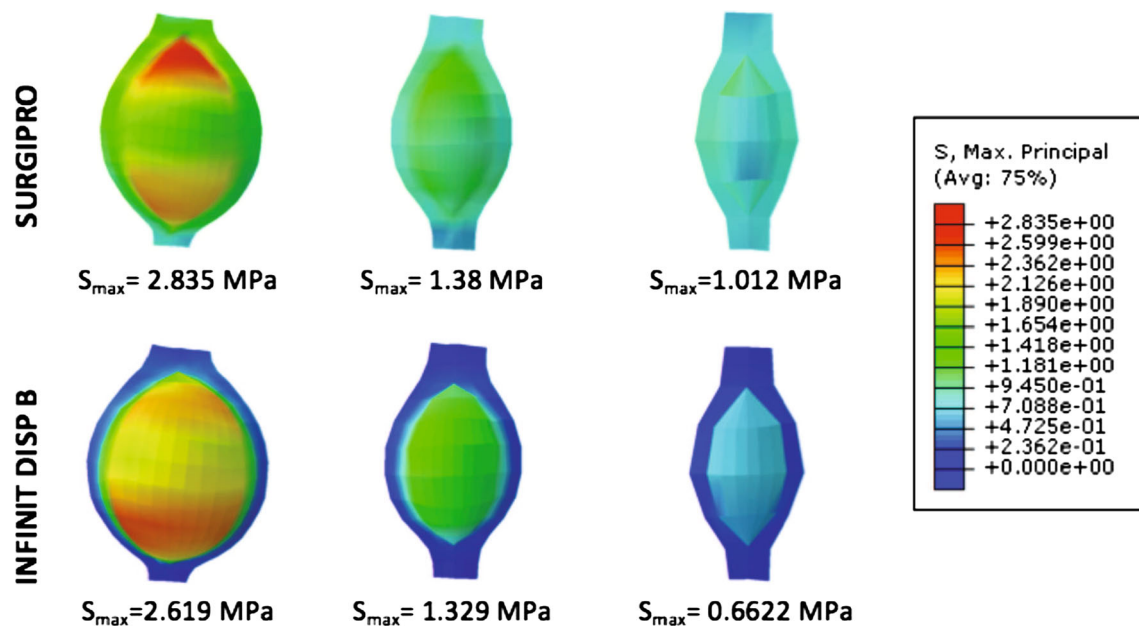
largest displacements in the prosthesis due to its compliance. The anisotropic behavior is more evident for the largest defect size. OPT and INF meshes behaved like an isotropic mesh for the small and the medium prostheses, since displacements distribution for the two dispositions remained quite similar, but they presented some differences for the large defect showing the anisotropic character. OPT mesh begins to have an anisotropic response for a stretch greater than 1.25 [12]. For the three meshes, the maximum value of displacements was not always reached at the center of the prosthesis, but rather it depended on the mesh stiffness. For INF mesh, due to its great compliance the central area is most vulnerable to suffer relaxation, so MD were reached in the middle of the prostheses whereas they remained quite low in the overlap zone. On the contrary, for the OPT mesh greater displacements were obtained in the overlap zone because this surgical mesh is stiffer than the tissue and restricted the movement. Only in large defect size MD were similar in the defect and the overlap. Concerning SUR mesh, higher displacements were also observed in the overlap zone for the three sizes.

Maximal principal cauchy stresses (MPS) along the line RL–LL are presented in Fig. 7. The evolution of the MPS along the path RL–LL shows a high increase of the stresses suffered by the prosthesis, due to the decrease of thickness in this area and the greater stiffness of the prostheses with respect to that of the tissue. This stress values increased with the size of the defect for all meshes. Along the line RL–LL, INF mesh showed higher MPS for large and medium size defect, followed by OPT mesh, while SUR remained with the lowest values of stress along the line. For small size, maximum values of stresses seemed to appear on the sides of the defect, where the prostheses joins to the overlap. However, this is difficult to distinguish due to the shortness of the segment. Attending to the anisotropy of the meshes, disposition A (with the stiffest direction of the mesh aligned with the transversal direction) presented

generally higher values of MPS, which was clearly visible in Fig. 7a for INF mesh. This difference between dispositions can be numerically seen in Table 1. In this table, maximum values of stretches and MPS recorded for the different prostheses, orientations and defect sizes are presented. Besides that, experimental breaking results (stretch and stress data) of the three synthetic meshes are also included. This breaking data corresponds to uniaxial tests performed on each surgical mesh, taken from Hernández-Gascón et al. [12]. According to the displacements produced in the prostheses the two extreme cases are SUR mesh and INF mesh disposition B, considered the stiffest and most compliant meshes, respectively. On that basis, the distribution of stresses over the defect for these two cases were obtained and plotted in Fig. 8. For INF mesh, higher values of stresses were obtained inside the prostheses area, while stresses in the overlap remained much lower. However, SUR mesh presented maximum values in the upper and lower zones close to the overlap for the large defect, and remained really close to values in the overlap for the other two sizes.

## Discussion

Numerical model are useful to study different surgical parameters. In this work we have analyzed the effect of key parameters, such as the mesh stiffness or its anisotropy, when the defect size is varied on ventral hernia repair in the immediate postoperative period. Based on a complex model of a healthy human abdomen with diverse anatomical structures differentiated, three different defect sizes were simulated to study its mechanical response to a high intraabdominal pressure (IAP) corresponding to a jumping motion (172.5 mmHg). Curvatures of the abdomen, displacements and stresses along the prostheses were calculated for all sizes and meshes here studied.



**Fig. 8** Distribution of maximal principal stresses (MPa) produced in the prostheses after a jumping motion

When we analyzed the response of the different prostheses, we found a general tendency for the three defect sizes: SUR mesh always led to the worse compliance of the abdominal wall, even restricting the movement of the healthy abdomen. This strong stiffening could lead to patient discomfort. On the other hand, OPT mesh appeared to have the same but slightly less effect for small and medium sizes while seemed to be closer to the natural distensibility of the abdomen for the large defect. Finally, INF mesh emerged as the most compliant mesh, exceeding in all cases the natural distensibility of the abdomen. According to these results, OPT mesh seemed to be the most suitable just after the implantation to avoid the possible discomfort generated in the patient by the SUR mesh or the excessive distension of the INF mesh that could lead in another hernia.

Concerning the influence of the mesh placement, we found that for large defect when the stiffest direction of the prosthesis gets aligned to the transversal direction (disposition A), it would lead to higher stresses and less displacements. These results agree with previous findings [22, 25] who suggested that the transversal direction of the abdomen is the stiffest one so the stiffness of the prostheses has to be considered before implantation. That is why, for OPT and INF meshes, disposition B is not the most appropriate mesh disposition for large defects. However, this difference in MD between A and B disposition was not that obvious for medium and small defects. As said before, OPT mesh begins to have an anisotropic response for a stretch greater than 1.25 [12], which could explain why below that value the prosthesis presented similar

deformation values. Regarding this aspect, both dispositions could be used for medium and small sizes if the slightly higher stress of disposition A can be supported by the prostheses.

The effect of bulging was present in the undeformed shape of the abdomen (with an IAP of 20 mmHg) for the three defect sizes and it increased after IAP was applied. INF mesh sharply showed this effect due to its great compliance and it enlarged with the size of the defect, which suggests bulging is strongly related with the size defect. Guérin and Turquier [16] with a simplified numerical model of the wall, come to the same conclusions.

Another factor to consider is the maximum stresses presented in each mesh. Numerical results, gathered in Table 1, indicates that SUR and OPT mesh did not exceed the breaking values for any defect size. However, INF mesh only could be used without breakage risk for small sizes in disposition A. Despite this isolate case we conclude that INF should not be used for high IAP, which agrees with the findings of Deeken et al. [26] who indicated that Ultrapro and Infinite meshes are not suitable for patients with obesity and large defects.

This study is not exempt from some limitations. First of all, due to the limited experimental data for human tissue characterization, numerical models were sometimes fed with animal data found in literature. Moreover, experimental data were derived from uniaxial tests in all cases, which could not be completely capable of reproducing multiaxial loading states. Other improvements required in this first approach to modeling hernia surgery include the consideration of viscoelastic effects, the loss of stiffness due



to damage after hernia appearance and the definition of a constitutive model that incorporates the formation of new collagenous tissue over the prosthesis as a response to the placement of a foreign body into the abdomen. Additionally, the study of the mechanics in the overlap should be widened. On one side, the join between the prostheses and the tissue has been simplified to a merger of nodes, and it should be deeply modeled. On the other side, when the mesh is mainly stretched in one direction, it tends to narrow in the perpendicular direction provoking an additional stress on the sutures that may affect to the overlap [27]. This effect has not been included in the current model and it may be considered for future works. Thus, further studies are needed to improve the understanding the mechanics of both, abdominal tissues and prostheses. Our simulations included data from *in vitro* testes, but need to be validated with other experimentation, such as *ex vivo* or *in vivo* tests. Some authors have conducted their research in this direction. Podwojewski et al. [28] for example, who performed *ex vivo* tests on human walls or Szymczak et al. [29], who described a protocol to study abdomen surface deformation due to some normal activities of people. More focused on the passive response of the abdomen is Song et al. [30], who used the internal pressure controlled by laparoscopy to inflate the abdomen and study its mechanical response. These experimental *ex vivo* or *in vivo* studies could be used to validate our simulations. To the best of our knowledge, there are no previous studies analyzing the effect of the prosthesis size in repaired herniated human abdomen.

To conclude, numerical simulations of hernia treatment could allow a better knowledge of the outcomes of hernia surgery. In addition, the methodology proposed could allow a better decision to choose the most appropriate prosthesis, in function of defect size, to improve the quality of patient life and prevent breakage of the mesh.

#### Compliance with ethical standards

**Conflict of interest** RSA declares no conflict of interest. BHG declares no conflict of interest. LL declares no conflict of interest. JMB declares no conflict of interest. EP declares no conflict of interest. BC declares no conflict of interest.

**Ethical approval** This article does not contain any studies with animals performed by any of the authors.

**Informed consent** Informed consent was obtained from all individual participants included in the study.

#### References

- Klinge U, Junge K, Stumpf M, Klosterhalfen B (2002) Functional and morphological evaluation of a low-weight monofilament polypropylene mesh for hernia repair. *J Biomed Mater Res B Appl Biomater* 63:129–136
- Klosterhalfen B, Junge K, Klinge U (2005) The lightweight and large porous mesh concept for hernia repair. *Expert Rev Med Devices* 2:103–117
- Brown SHM, McGill SM (2010) A comparison of ultrasound and electromyography measures of force and activation to examine the mechanics of abdominal wall contraction. *Clin Biomech* 25:115–123
- Cobb WS, Burns JM, Kercher KW, Matthews BD, Norton HJ, Heniford BT (2005) Normal intraabdominal pressure in healthy adults. *J Surg Res* 129:231–235
- Langer C, Neufang T, Kley C, Liersch T, Becker H (2001) Central mesh recurrence after incisional hernia repair with marlex— are the meshes strong enough? *Hernia* 5:164–167
- Lintin LAD, Kingsnorth AN (2014) Mechanical failure of a lightweight polypropylene mesh. *Hernia* 18:131–133
- Blázquez Hernando LA, García Ureña MA, López Monclús J, Lersundi Robin del Valle A, Melero Montes D, Cruz Cidoncha A, Jiménez Ceinos C, Castellón Pavón C (2015) Roturas de malla: una causa poco frecuente de recidiva herniaria. *Rev Hispanoam Hernia* 3(4):155–159
- Zuvela M, Galun D, Djuric-Stefanovic A, Palibrk I, Petrovic M, Milicevic M (2014) Central rupture and bulging of low-weight polypropylene mesh following recurrent incisional sublay hernioplasty. *Hernia* 18:135–140
- Schippers E (2007) Recurrent Hernia: Prevention and Treatment, chapter Central mesh rupture—Myth or real concern? Springer, Berlin, Heidelberg, pp 371–376
- Hernández-Gascón B, Peña E, Grasa J, Pascual G, Bellón JM, Calvo B (2013) Mechanical response of the herniated human abdomen to the placement of different prostheses. *J Biomech Eng* 135(5):51004
- Anurov MV, Titkova SM, Oettinger AP (2010) Impact of Position of Light Mesh Endoprosthesis with Anisotropic Structure for the Efficiency of Anterior Abdominal Wall Reconstruction. *Bull Exp Biol Med* 149(6):779–783
- Hernández-Gascón B, Peña E, Melero H, Pascual G, Doblaré M, Ginebra MP, Bellón JM, Calvo B (2011) Mechanical behaviour of synthetic surgical meshes. Finite element simulation of the herniated abdominal wall. *Acta Biomater* 7:3905–3913
- Ozog Y, Konstantinovic ML, Werbrouck E, De Ridder D, Mazza E, Deprest J (2011) Persistence of polypropylene mesh anisotropy after implantation: an experimental study. *Urogynaecology* 118(10):1180–1185
- López-Cano M, Rodríguez-Navarro J, Rodríguez-Baeza A, Armengol-Carrasco M, Susín A (2007) A real-time dynamic 3D model of the human inguinal region for surgical education. *Comput Biol Med* 37:1321–1326
- Fortuny G, Rodríguez-Navarro J, Susín A, López-Cano M (2009) Simulation and study of the behaviour of the transversalis fascia in protecting against the genesis of inguinal hernias. *J Biomech* 42:2263–2267
- Guérin G, Turquier F (2013) Impact of the defect size, the mesh overlap and the fixation depth on ventral hernia repairs: a combined experimental and numerical approach. *Hernia* 17:647–655
- Hernández-Gascón B, Mena A, Peña E, Pascual G, Bellón JM, Calvo B (2012) Understanding the passive mechanical behavior of the human abdominal wall. *Ann Biomed Eng* 41(2):433–444
- Gerovichev O, Marayong P, Okamura AM (2002) The effect of visual and haptic feedback on manual and teleoperated needle insertion. In: *Medical Image Computing and Computer-Assisted Intervention—MICCAI*, vol 2488. Springer, Berlin, Heidelberg, pp 147–154
- Hernández-Gascón B, Peña E, Pascual G, Bellón JM, Calvo B (2014) Computational modeling of objects presented in images: fundamentals, methods and applications, chapter can numerical

- modelling help surgeons in abdominal Hernia Surgery? Springer International Publishing, New York, pp 167–185
20. Sabbagh C, Dumont F, Robert B, Badaoui R, Verhaeghe P, Regimbeau JM (2011) Peritoneal volume is predictive of tension-free fascia closure of large incisional hernias with loss of domain: a prospective study. *Hernia*, pp 559–565
  21. Moore W (2008) Gray's Anatomy celebrates 150th anniversary. *The Telegraph* (Telegraph Media Group)
  22. Hernández B, Peña E, Pascual G, Rodríguez M, Calvo B, Doblare M, Bellón JM (2011) Mechanical and histological characterization of the abdominal muscle. A previous step to model Hernia surgery. *J Mech Behav Biomed* 4:392–404
  23. Martins P, Peña RM, Natal Jorge RM, Santos A, Santos L, Mascarenhas T, Calvo B (2012) Mechanical characterization and constitutive modelling of the damage process in rectus sheath. *J Mech Behav Biomed* 8:111–122
  24. Schoenmaeckers EJ, Wassenaar EB, Raymakers JT, Rakic S (2010) Bulging of the mesh after laparoscopic repair of ventral and incisional hernias. *JSLs* 14(4):541–546
  25. Morrow DA, Donahue TLH, Odegard GM, Haufman KR (2010) Transversely isotropic tensile material properties of skeletal muscle tissue. *J Mech Behav Biomed Mater* 3:124–129
  26. Deeken CR, Abdo MS, Frisella MM, Matthews BD (2011) Physicomechanical evaluation of polypropylene, polyester, and polytetrafluoroethylene meshes for inguinal hernia repair. *J Am Coll Surg* 212(1):68–79
  27. Binnebösel M, Rosch R, Junge K, Flanagan TC, Schwab R, Schumpelick V, Klinge U (2007) Biomechanical analyses of overlap and mesh dislocation in an incisional hernia model in vitro. *Surgery* 142(3):365–371
  28. Podwojewski F, Otténio M, Beillas P, Guérin G, Turquier F, Mitton D (2014) Mechanical response of human abdominal walls ex vivo: Effect of an incisional hernia and a mesh repair. *J Mech Behav Biomed Mater* 38:126–133
  29. Szymczak C, Lubowiecka I, Tomaszewska A, Smietański M (2012) Investigation of abdomen surface deformation due to life excitation: implications for implant selection and orientation in laparoscopic ventral hernia repair. *Clin Biomech (Bristol, Avon)* 27(2):105–110
  30. Song C, Alijani A, Frank T, Hanna G, Cuschieri A (2006) Mechanical properties of the human abdominal wall measured in vivo during insufflation for laparoscopic surgery. *Surg Endosc* 20:987–990