Combined in vivo and ex vivo analysis of mesh mechanics in a porcine hernia model

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Abstract

Background Hernia meshes exhibit variability in mechanical properties, and their mechanical match to tissue has not been comprehensively studied. We used an innovative imaging model of in vivo strain tracking and ex vivo mechanical analysis to assess effects of mesh properties on repaired abdominal walls in a porcine model. We hypothesized that meshes with dissimilar mechanical properties compared to native tissue would alter abdominal wall mechanics more than better-matched meshes.

Methods Seven mini-pigs underwent ventral hernia creation and subsequent open repair with one of two heavyweight polypropylene meshes. Following mesh implantation with attached radio-opaque beads, fluoroscopic images were taken at insufflation pressures from 5 to 30 mmHg on postoperative days 0, 7, and 28. At 28 days, animals were euthanized and ex vivo mechanical testing performed on full-thickness samples across repaired abdominal walls. Testing was conducted on 13 mini-pig controls, and on meshes separately. Stiffness and anisotropy (the ratio of stiffness in the transverse versus craniocaudal directions) were assessed.

Results 3D reconstructions of repaired abdominal walls showed stretch patterns. As pressure increased, both meshes expanded, with no differences between groups. Over time, meshes contracted 17.65% (Mesh A) and 0.12% (Mesh B; \( p = 0.06 \)). Mesh mechanics showed that Mesh A deviated from anisotropic native tissue more than Mesh B. Compared to native tissue, Mesh A was stiffer both transversely and craniocaudally. Explanted repaired abdominal walls of both treatment groups were stiffer than native tissue. Repaired tissue became less anisotropic over time, as mesh properties prevailed over native abdominal wall properties.

Conclusions This technique assessed 3D stretch at the mesh level in vivo in a porcine model. While the abdominal wall expanded, mesh-ingrown areas contracted, potentially indicating stresses at mesh edges. Ex vivo mechanics demonstrate that repaired tissue adopts mesh properties, suggesting that a better-matched mesh could reduce changes to abdominal wall mechanics.

Keywords Ventral hernia · Mesh · Porcine model · Mechanics

There are over 350,000 ventral hernia repairs performed annually in the US, making this one of the more common general surgery procedures performed [1]. While utilizing mesh in hernia repair has improved surgical outcomes by reducing recurrence rates from 63 to 32% over primary suture repair [2], the use of mesh is not without potential morbidity [2–5]. Current outcomes reveal a need to expand our understanding of how meshes affect the abdominal wall mechanics.
wall in order to mitigate complications and reduce recurrence rates.

Meshtes are often designed to be stronger than the abdominal wall [6]. While appropriate tensile strength is an important mechanical parameter in mesh design, the majority of hernia recurrences occur at the mesh edges [7]. Recurrences have been shown to be caused by several factors, including insufficient mesh overlap [7–9], myofascial layer closure failure [7–9], damage from fixation devices [9, 10], wound contracture [5, 11], poor tissue ingrowth [3–5], and sustained inflammatory response [3–5, 12]. However, mechanical mismatch between the mesh and ingrown tissue has not been fully evaluated.

The concept of biomimicry, the design of devices to match native anatomic function, has not permeated hernia mesh design from the mechanical standpoint. The comparison between native tissue and non-absorbable synthetic meshes has been studied, and uniaxial tensile properties of explanted meshes showed histological and mechanical changes post-repair. This includes an increase in collagen growth after the repair, causing mechanical stiffness changes [13]. In a mathematical model of abdominal wall behavior post-implantation, it was found that the mesh which most closely matched the native tissue properties resulted in less changes to the abdominal wall mechanics than the mesh that did not closely match native properties [14]. These studies, while forming the idea of advancing hernia mesh design towards biomimicry, have yet to comprehensively demonstrate the importance of mechanics in this process.

While these previous studies have provided some insight, no approach has been able to fully evaluate mechanics of a mesh-implanted abdominal wall. A better understanding of this relationship could influence the development of more effective hernia meshes and improve the success of hernia repair. Additionally, this could potentially help push the paradigm of synthetic implant design towards mechanical biomimicry. The repaired area at the subsurface level of the mesh has not been studied in both in vivo and ex vivo settings to assess spatial and temporal changes caused by synthetic mesh implantation. Due to limitations in studying mechanical mesh properties in humans, we sought to determine the degree to which mesh-tissue mechanical match drives integration and function at the subsurface level of the mesh in a porcine model of ventral hernia repair. Specifically, through the utilization of novel in vivo 3D strain tracking paired with ex vivo biaxial evaluation of location-specific mechanical properties, we aimed to determine whether a mechanical mismatch between the abdominal wall and the implanted mesh leads to altered mechanical properties.

Materials and methods

Porcine model

We utilized a previously established porcine model of ventral hernia repair that has been used extensively to evaluate tissue remodeling of a wide range of biomaterials [4, 15–19]. Yucatan mini-pigs were chosen due to the clinically relevant size of the abdomen and slow growth, allowing for implantation of mesh with dimensions similar to those utilized clinically. In addition, the porcine abdomen is similar to that of a human in terms of musculature and strength.

A total of 20 Yucatan mini-pigs with an average initial weight of 56 ± 7.4 lbs were utilized in this study: 13 male pigs were used as a native tissue control group for ex vivo mechanical testing, and seven female Yucatan pigs were used for the treatment group, which including the creation of ventral midline hernias with subsequent mesh repair. Differences in sex were due to availability of control animals. A total of ten treatment group pigs underwent hernia repair using one of two synthetic meshes, but three were rejected from data analysis after day 0 due to postoperative infection.

Surgical approach

This study was performed under strict compliance with a protocol approved by the Animal Studies Committee of Washington University School of Medicine, and all procedures were performed under sterile conditions. For the treatment group (n = 7), hernias were created by making a longitudinal midline incision of 5 cm that was 1 cm superior to the umbilicus. The incision was extended through the skin, subcutaneous fat, and fascia, but not through the peritoneum. The fascia was left open, but the subcutaneous tissue and skin was closed in layers. The abdominal wall defect was left untreated for 21 days to achieve a mature hernia. After maturation, the previously created wound was opened down to the underlying peritoneum. A retromuscular plane was developed bilaterally out to the linea semilunaris and the posterior rectus sheath was closed in the midline. The evaluated meshes include Mesh A (n = 4) and Mesh B (n = 3), both commercially available heavyweight, permanent, non-coated polypropylene meshes. Mesh A had a weight of 109 g/m² and a pore area of 0.39 mm², and Mesh B had a weight of 105 g/m² and a pore area of 0.44 mm². Meshes were cut to 10 × 10 cm and positioned in the posterior sheath with the stiffer direction of the mesh aligned transversely to match anatomic stiffness. Meshes were selected based on previous biaxial mechanical analysis, which revealed that Mesh
B represented a closer match to native porcine abdominal wall mechanics in terms of stiffness and directional differences, while Mesh A represented a poorer match [20]. Specifically, Mesh A was stiffer than Mesh B. Mesh A was also found to be isotropic, or exhibiting the same stiffness in both directions, whereas Mesh B was anisotropic.

A grid of radio-opaque stainless steel beads was sutured to each mesh prior to implantation (Fig. 1). In addition, eight radio-opaque tacks were fired 1–2 cm peripheral to the mesh in the abdominal wall intra-operatively (Fig. 2). These radio-opaque markers were used for in vivo strain tracking, as described below. Eight transabdominal fixation sutures were used to secure the mesh in place. The fascia and skin were closed as previously described.

**Fluoroscopic image acquisition**

Immediately following hernia repair, fluoroscopy equipment and calibration frames (Fig. 3) were moved into position. Intra-abdominal pressure (IAP) was then incrementally increased from 5 to 30 mmHg in 5 mmHg steps. At each pressure, pairs of fluoroscopic images were acquired with the C-arm fluoroscope positioned ±15° relative to the vertical axis [21]. Changes in IAP were used to replicate in vivo forces. To prevent abdominal movement associated with respiration from impacting tack positions, a temporary breath-hold was induced while the C-arm fluoroscope rotated between the two angular positions and image pairs were captured. Analysis techniques included custom MATLAB (The MathWorks Inc., Natick, MA) programs to reconstruct pairs of 2D images in 3D, as previously described [21]. In vivo positions of radio-opaque beads were determined and used to compute 3D strain profiles at each incremental step of IAP. A strain tracking program similar to what was previously developed computed areal stretch of the abdominal wall as functions of time and pressure [22]. Mesh area was calculated using a MATLAB program that uses in vivo bead centroid positions to compute the cross product of triangles formed by adjacent beads, and thus the 3D mesh area [23].

**Evaluation throughout healing**

At 7 and 28 days post-repair, the animals were sedated and the abdomen was insufflated under sterile conditions through access with a Veress needle. Paired X-ray images were again acquired and analyzed at each pressure (from 5 to 30 mmHg in 5 mmHg increments) to determine how the mechanics of the abdominal wall changed at early and mid-stages of healing following hernia repair. Areal stretch was
examined within the mesh-repaired area (by tracking the 16 bead radio-opaque grid on the mesh) and also in the peri-implant tissues (by tracking radio-opaque tacks deployed into neighboring tissues; Fig. 2).

**Tissue procurement and biomechanical testing**

Animals were euthanized at 28 days post-repair, and test specimens were procured from specific locations of the porcine abdomen, namely mesh-tissue complexes from each side of the midline, and tissue-only samples from locations lateral and caudal to the repaired hernia defect (Fig. 2). Skin and subcutaneous fat were removed from samples, and all fascial and muscle layers were left intact. The removal of skin and subcutaneous fat led to a more clinically relevant testing method since it has been previously shown that internal and external strains can be significantly different in the abdominal wall [24, 25]. In order to test underlying tissue health, where hernias occur and mesh is placed, isolating the mechanics of subsurface layers is more critical than understanding skin and fat mechanics. Mesh-tissue complexes and caudal samples were composed of only rectus muscle, while lateral samples were part of the rectus complex and included both rectus and transversus abdominis muscles. Caudal samples were procured from below the arcuate line (BAL), and therefore contained no posterior rectus sheath. Control animals were euthanized based on availability without hernia creation or repair, and samples were harvested from the same locations.

Square (3.5 × 3.5 cm) full-thickness samples were tested within 16 h following animal euthanasia. Samples were marked to maintain proper orientation, thicknesses were measured using a non-contact laser scanning system, and tensile biaxial mechanical testing was performed in a phosphate-buffered saline (PBS) bath to maintain hydration. Mesh A \( n = 5 \), Mesh B \( n = 5 \), and native full-thickness (FT) samples from 13 control animals \( n = 36 \) were mechanically tested as controls.

Linear stiffness and anisotropy (ratio of the transverse stiffness to craniocaudal stiffness) were evaluated for native tissue, repaired tissue, and meshes alone. Treatment groups in the mesh-overlap region included Mesh A-Tissue Complex (A-TC) and Mesh B-Tissue Complex (B-TC).

**Data analysis**

In vivo fluoroscopic data were visualized to assess differences in areal stretch. \( T \) tests were utilized to determine differences between meshes in terms of contracture and expansion throughout insufflation and healing.

Ex vivo mechanical parameters were analyzed using custom-written MATLAB programs. Linear stiffness (N/mm) was computed by using least squares regression to find the slope of the linear region of the force–displacement curves. Anisotropy, the ratio of transverse to craniocaudal stiffness values, was also computed. Anisotropy is a measure of directional dependence. A value of 1, or isotropy, indicates that a material is as stiff transversely as it is craniocaudally. Samples were statistically compared using ANOVA tests across locations, mesh types, and against native tissue control samples. Statistical significance was set at \( p < 0.05 \).

**Results**

**In vivo fluoroscopic analysis**

3D reconstructions of the repaired abdominal wall showed areal stretch as functions of pressure and time (Fig. 4).
Both mesh beads and tacks placed peripheral to the mesh were analyzed and visualized. An areal stretch of 1 (i.e., no change from the reference state) is shown in white on the color bar, while red and blue show expansion and contraction, respectively. General trends show that meshes expanded with increasing pressure and contracted over time following surgical repair.

Initial calculated mesh area was 88.5 cm² for Mesh A-implanted animals and 86.5 cm² in Mesh B-implanted animals. This area represents the surface enclosed by the centroids of the 16 beads sutured onto the mesh, and differed slightly due to the precise location of bead placements on the mesh. As the abdomen was insufflated, both meshes expanded, with no differences between groups. Mesh A-implanted mesh areas expanded 3.3% from 5 to 30 mmHg on day 0, and Mesh B-implanted mesh areas expanded 3.1% (Fig. 5).

Over time, Mesh A-implanted meshes contracted an average of 17.65%, while Mesh B-implanted meshes contracted 0.12% ($p = 0.06$). Peripheral abdominal wall area, or the area enclosed by the space between the peripheral abdominal wall tacks and the mesh edges, expanded with animal growth, and was not significantly different between groups (Fig. 6). Visualizations of mesh and peripheral areal changes over time showed qualitative differences between Mesh A-implanted meshes and Mesh B-implanted meshes (Fig. 7). Specifically, Mesh A contracted uniformly across the surface of the mesh, whereas the pattern of stretch for Mesh B varied across the mesh. Both meshes appeared to contract more craniocaudally than transversely, and contraction at the linea alba appeared to be greater than on the lateral edges.

**Fig. 4** Undeformed implanted mesh on day 0 (left) expands with pressure (top right) and contracts with time (bottom right). An areal stretch of 1 indicates no change.

**Fig. 5** Mesh A- and Mesh B-implanted mesh areas expand as a function of pressure on day 0. No differences were seen between mesh groups.

**Fig. 6** Mesh A- and Mesh B-implanted mesh behavior over time. The peripheral area expands with time and no statistical differences were seen in mesh contracture at day 28 between groups ($p = 0.06$).
Ex vivo biaxial biomechanical evaluation

Mesh B matched the full-thickness (FT) rectus region of the native abdominal wall better in terms of linear stiffness. In both craniocaudal and transverse directions, Mesh A alone was stiffer than native rectus FT \( (p < 0.05) \). Cranio-caudally, it was more than twice as stiff \( (16.21 \text{ vs. } 7.16 \text{ N/mm}) \). Mesh B alone did not differ from the native rectus FT in either direction (Table 1).

Native FT tissue was also evaluated across the abdominal wall, including lateral and below the arcuate line (BAL) regions in both the craniocaudal and transverse directions (Fig. 8). Native FT tissue was stiffer in the transverse direction than the craniocaudal direction in all regions. The rectus region was stiffest \( (7.16 \text{ N/mm transversely}) \), and BAL was least stiff \( (3.76 \text{ N/mm transversely}) \).

Compared to native tissue, A-TC and B-TC were stiffer in the mesh-overlap region in both directions (Fig. 8; \( p < 0.05 \)). The greatest differences were seen in the craniocaudal direction, where native tissue was 7.16 N/mm and A-TC was 20.67 N/mm. B-TC, while also significantly different from native tissue, was 14.35 N/mm. This shows a stepwise increase in stiffness from native tissue to B-TC and then A-TC, which matches what was seen in the mesh properties when they were tested independently (Table 1).

The thickness of the rectus region of native tissue was \( 8.80 \pm 1.71 \text{ mm} \), whereas A-TC had a thickness of \( 14.31 \pm 1.77 \text{ mm} \) \( (p < 0.05) \) and B-TC was \( 14.66 \pm 1.93 \text{ mm} \) (Table 2; \( p < 0.05 \)).
When treatment group specimens from tissue-only areas peripheral to the mesh were compared to native tissue from equivalent regions (i.e., lateral region of A-TC compared to lateral region of native tissue), no differences were seen in either lateral or BAL regions. A-TC and B-TC exhibited different BAL stiffness values in the transverse direction than each other, but neither was significantly different from native BAL tissue (Fig. 8).

The anisotropy ratio of native rectus FT (2.06 ± 0.55) was compared to meshes alone, repaired tissue, and other regions of native tissue (Fig. 9). All groups except native BAL FT were significantly different than native FT tissue in the rectus region. Mechanics of the meshes alone showed that Mesh A deviated from anisotropic native rectus tissue more than Mesh B. The anisotropy ratio for native rectus FT was 2.06 ± 0.55, while Mesh B and Mesh A had more isotropic ratios of 1.52 ± 0.27 and 1.11 ± 0.20, respectively. Both mesh-implanted groups became more isotropic over time as mesh properties prevailed over native abdominal wall properties. A-TC and B-TC followed anisotropy ratios in a stepwise pattern, indicating that the directional dependence of ingrown tissue appears to be dependent upon mesh properties. No difference was found between native tissue and each tissue complex. The only area with differences outside of the mesh/rectus region was below the arcuate line (BAL). A-TC and B-TC differed in transverse stiffness, but neither was different than native tissue.

Table 1 Linear stiffness values for native tissue and meshes

<table>
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<th>Craniocaudal stiffness (N/mm)</th>
<th>Transverse stiffness (N/mm)</th>
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<tbody>
<tr>
<td>Native rectus FT</td>
<td>7.16 ± 2.11</td>
<td>13.74 ± 3.14</td>
</tr>
<tr>
<td>Mesh A</td>
<td>16.21 ± 1.15*</td>
<td>18.10 ± 1.75*</td>
</tr>
<tr>
<td>Mesh B</td>
<td>9.25 ± 1.79</td>
<td>12.70 ± 1.63</td>
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Native rectus FT was full-thickness (FT) tissue from the rectus region of 13 animals (n = 44). Meshes were tested independently of tissue. Differences were found between Mesh A (n = 5) and Native rectus FT (*p < 0.05) in both directions, and no differences were found between Mesh B (n = 5) and Native rectus FT.

Table 2 Thickness of tissue as measured by a non-contact laser scanner

<table>
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<th>Rectus region thickness (mm)</th>
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<tbody>
<tr>
<td>Native rectus FT</td>
<td>8.80 ± 1.71</td>
</tr>
<tr>
<td>A-TC</td>
<td>14.31 ± 1.77*</td>
</tr>
<tr>
<td>B-TC</td>
<td>14.66 ± 1.93*</td>
</tr>
</tbody>
</table>

Both treatment groups, represented by mesh-tissue complexes, were thicker than native rectus full-thickness samples (*p < 0.05).

When treatment group specimens from tissue-only areas peripheral to the mesh were compared to native tissue from equivalent regions (i.e., lateral region of A-TC compared to lateral region of native tissue), no differences were seen in either lateral or BAL regions. A-TC and B-TC exhibited different BAL stiffness values in the transverse direction than each other, but neither was significantly different from native BAL tissue (Fig. 8).

Discussion

Hernia meshes often exhibit a higher stiffness or strength compared to native abdominal wall tissue. The relationship between these properties and hernia repair complications...
remains unknown. Previous techniques have never been able to elucidate behavior at the level of the mesh in vivo while also studying location-specific mechanics ex vivo. To date, no study has successfully gone beyond using theoretical models [26–30], tracking surface strains in vivo [31–40], or investigating mechanical properties of isolated tissue samples ex vivo [13, 24, 27, 28, 30, 41–51] to understand the effects of the mesh mechanics on the abdominal wall tissue. Several studies have also used X-ray [52], computed tomography (CT) [53], and MRI [1, 54, 55] to evaluate post-repair abdominal walls to understand mesh positioning over time. While mesh contracture has been highly reported [54, 56], in these models it was not possible to study mechanics at several time points both in vivo and ex vivo. Through a novel use of biplanar fluoroscopy, the mechanics of hernia meshes post-implantation were studied across time and pressure. This comprehensive analysis permitted a deeper understanding of mesh behavior, which is critical for advancing biomimicry in mesh design, possibly leading to lower recurrence rates.

In the present study, there were no differences between groups in immediate postoperative expansion with insufflation. While the meshes exhibited different mechanical properties at baseline and after explantation, as seen by the biaxial mechanical analysis, these effects were not elucidated in vivo as a function of pressure. This finding suggests that in the immediate recovery period of hernia repair, the mesh mechanics are less critical than other factors in altering the behavior of the abdominal wall. Individual animals behaved differently, highlighting a large variance in the data, and suggesting the need to increase sample size in further studies.

Potential differences were in seen in areal stretch over time, discussed as mesh contracture and expansion, which potentially indicates the creation of stress concentrations at the mesh edges. Since the peripheral area of the abdominal wall was not found to differ between groups over time while the mesh area changed, the stretch patterns were affected. This was visually confirmed through the areal stretch fluoroscopic reconstructions. With the expansion of the wall and contraction of the mesh-overlap area, these patterns over a prolonged period of time could lead to weaker tissue at the mesh edge. Clinically, central mesh failure is a rare mechanism of recurrence due to the strong nature of meshes [6, 7]. Repairs more frequently fail at the mesh edge, where overlap or fixation may not be enough due to inflammation or mesh contracture [7–9]. The cause of mesh contracture has not been adequately studied in literature. While it has been shown that a reduction in amount of polypropylene material can lead to less shrinkage or contracture [11], this is the first study to highlight the potential correlation between mechanical properties and contracture. These results indicate that a mesh that has better-matched stiffness values might reduce mesh contracture, potentially mitigating additional stresses.

Biaxial mechanical analysis revealed that both Mesh A and Mesh B are different than native FT tissue in the rectus area. A-TC and B-TC followed the mechanics of the meshes alone, indicating that repaired tissue adopts properties more similar to the implanted mesh. When paired with the contracture and potential stress concentrations exhibited by implanted Mesh A over time, it makes a strong case for advancing the concept of biomimicry in hernia mesh design. By implanting a mesh that matches the desired area of the abdominal wall in terms of stiffness and anisotropy, it is possible that this would mitigate changes to the mechanics of the overlap region.

Abdominal wall mechanics were found to be location-dependent, which has yet to be considered in mesh design or placement. Mesh A matches the anisotropy of only one region, and is stiffer than native tissue in all regions. Mesh B better matches the stiffness of both the craniocaudal and transverse direction of the rectus region, and exhibits an anisotropy ratio in between the ratios of the rectus and BAL regions. These parameters should be emphasized and studied in mesh design to better match all (or specific) parts of the abdominal wall. Larger meshes often overlap several areas of the abdominal wall, and current mesh designs exhibit the same properties throughout, creating a mismatch in parameters across the abdominal wall. Orientation

![Fig. 9](image_url) The anisotropy ratio or a measure of directional dependence by taking the ratio of transverse stiffness to craniocaudal stiffness. A value of 1 indicates isotropy, or the same properties in both directions. Statistical significance (*) was measured against Native rectus FT and was seen in all meshes and tissue complexes. Native lateral FT was also different than Native rectus FT, showing differences across the abdominal wall.
when placing the mesh should also be considered at the
time of implantation, as some meshes exhibit significant
anisotropy [20, 57].

The small sample size is a limitation to the quantitative
nature of the fluoroscopy in this study, in part due to
infections in some of the animals, which were removed
from data analysis after the initial day 0 fluoroscopy. The
source of infection was unclear, but there is no reason to
believe the in vivo analysis caused complications, or that
the uninfected animals were compromised. The novel
biplanar fluoroscopic system has been validated [21], and
further studies are needed to increase the statistical power
to detect differences.

While we noted that Mesh A trends towards contracting
more than Mesh B in vivo, the ex vivo results suggest a
stronger quantitative analysis of different mechanical
behavior between meshes. The ex vivo mechanics achieves
a higher resolution by testing smaller samples, thus yielding
a greater sample size and more statistical power. The
in vivo fluoroscopy, on the other hand, is limited by the
number of tasks, and expansion and contraction was
measured across the entire mesh surface. While in vivo
data may provide greater clinical insight, this additional
level of variability was controlled for in the ex vivo testing.

There was also a large variability between animals, and
several factors presented challenges in addressing this
variability. It is possible that some meshes were initially
implanted with greater tension, potentially changing the
strain as the pressure was varied. Breath holds were initi-
ated to maintain constant abdominal wall position during
the imaging procedure, but both the C-arm fluoroscope and
the pig were susceptible to minor movement in between
acquisition of paired images. This could have potentially
created accuracy issues between the two images that were
used to create 3D reconstructions of the abdominal wall.

While the Yucatan mini-pig abdominal wall is similar in
size to that of a human, clinically relevant force exertions,
such as coughing or vomiting, were unable to be replicated
beyond abdominal cavity insufflation. The implanted mesh
also caused an inflammatory response that significantly
increased the thickness of the tissue in the overlap region.
This increased thickness, a change potentially just seen in a
porcine model, could have altered tissue mechanics and
compounded the changes.

This study has analyzed the relationship between mesh
mechanical properties and tissue complexes post-implan-
tation, in addition to highlighting the potential impact of
mesh properties on strain in vivo across time and pressure.
While the clinical significance of mesh contracture and
changing tissue mechanics is still relatively unknown, this
study provides a framework for advancing the under-
standing of how biomimicry could impact hernia mesh
design. By designing meshes that better match the location-
dependent mechanics of the native abdominal wall, changes
in mechanics could be mitigated and meshes could
potentially contract less over time; the reduction of stress
concentrations at mesh edges could decrease recurrences.

Supplemental material

Mechanical testing methods

Each sample was mounted on a planar biaxial test machine
(TestResources, Shakopee, MN) with a set of five needles
passed through slotted custom fixtures. After applying a
0.2 N pre-load on each actuator, samples were subjected to a
multi-step loading protocol. First, samples underwent
equibiaxial cyclic loading using a triangular waveform to
10% clamp–clamp displacement at 0.1 Hz for 10 cycles
(Step 1), followed by two strip tests, where one axis remained
fixed in location, while the other was loaded cyclically under
the same conditions as Step 1. The craniocaudal oriented axis
was cycled first with the transverse axis remaining fixed
(Step 2), and then the transverse axis was cycled with cran-
iocaudal axis fixed (Step 3). Finally, the samples were
ramped biaxially at 0.1 mm/s until signs of failure became
evident (Step 4). A similar protocol was used to characterize
synthetic meshes by our group previously [20].

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