

Recent developments in advanced manufacturing technologies for hernia repair meshes

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Received: 17 April 2026 / Accepted: 25 September 2026

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Abstract

Hernias are one of the most common pathologies, with over 20 million surgeries performed annually worldwide. The most commonly used surgery to treat this pathology involves the implantation of fibrous mesh structures to reinforce weakened abdominal tissue. Although commercially available synthetic and biological meshes have significantly reduced recurrence rates, they are still associated with postoperative complications, such as chronic inflammation, foreign body response, and intra-abdominal adhesions. This review focuses on the use of advanced manufacturing techniques, specifically extrusion-based three-dimensional (3D) printing and electrospinning, in the development of next-generation mesh structures for hernia repair. These techniques offer precise control over structural architecture, customizable mechanical properties, and the ability to integrate bioactive compounds for patient-specific implant design. Furthermore, the combination of extrusion-based 3D printing and electrospinning presents a synergistic approach, utilizing 3D printing for robust mechanical support and electrospinning to mimic the native extracellular matrix (ECM) and regulate the biological interface, ultimately aiming to optimize tissue integration and minimize postoperative complications. In addition to the critical analysis of current advances, this review aims to serve as a practical guide for researchers in this field, providing a structured overview of the reported key parameters and conditions.

Keywords Electrospinning · 3D printing · Hernia mesh · Multilayered scaffolds · Bioactive mesh

1 Introduction

Hernias are a common medical condition in which organs or tissues abnormally protrude through an opening or weakness in the surrounding muscle or connective tissue. Although they can occur in various parts of the body, abdominal hernias are particularly frequent. It is estimated that more than 20 million hernia repair surgeries are performed each year globally, reflecting not only the high incidence of this condition but also the clinical, economic, and social impact associated with its treatment [1]. These defects are driven by intrinsic genetic vulnerabilities, congenital abnormalities, or extrinsic factors that exert pressure on the abdominal cavity, such as chronic

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coughing, obesity, and intense physical stress [2]. To treat these defects and prevent recurrence, surgical meshes were developed to provide tension-free reinforcement and support tissue growth.

Currently, the clinical standard involves three main categories of meshes: synthetic non-absorbable, synthetic absorbable, and biological meshes. While non-absorbable synthetic meshes are widely used for their high mechanical strength and durability, they often induce chronic inflammatory reactions, foreign body responses, and visceral adhesions [3]. Conversely, absorbable meshes may degrade before sufficient tissue stability is achieved, and biological meshes, while promoting tissue regeneration with lower inflammatory responses, are limited by high costs, lower durability, and a tendency to creep over time [4, 5]. Composite meshes, which combine mechanically robust support with absorbable, biological, or functional layers, therefore represent a particularly promising route for overcoming the limitations of single-material systems and developing more balanced solutions for hernia repair.

Due to these limitations, a need for new manufacturing strategies has emerged to develop more advanced, personalized implants. In recent years, advanced manufacturing techniques have attracted increasing attention because they overcome the design restrictions of traditional manufacturing methods. These emerging technologies, specifically 3D printing and electrospinning, allow precise control over mesh architecture, from the micro to the macroscale [6, 7]. 3D printing, particularly through FDM, enables the creation of robust mechanical structures with precise control over their architecture [8]. On the other hand, electrospinning enables the production of fibers with diameters ranging from micrometers to nanometers, resulting in membranes with a high surface-to-volume ratio and interconnected porosity that mimic the structure of the native ECM, thereby promoting cell adhesion and proliferation [5, 7, 9].

The synergistic combination of these two techniques emerges as a particularly promising approach for the development of next-generation meshes [7]. This hybrid strategy aims to unite the mechanical support provided by 3D printing with the biofunctionality of the biological interface offered by electrospinning, optimizing tissue integration and minimizing postoperative complications, such as adhesions and chronic inflammatory response [10].

In this context, this review aims to provide a critical analysis of commercially available hernia meshes and recent studies on the use of advanced manufacturing techniques in the production of fibrous mesh structures for hernia treatment. Specifically, this article focuses on FDM 3D printing and electrospinning approaches, comparing their respective advantages and limitations with those of commercially available meshes, and highlighting current research trends, challenges, and future prospects. The literature search was conducted using the Web of Science database with the keyword combinations “hernia mesh AND 3D printing” and “hernia mesh AND electrospinning”, focusing on studies published over the last ten years (2017–2025). The relevant studies were selected based on their contribution to the production, characterization, or biological evaluation of mesh structures. Although other 3D printing technologies may offer specific advantages, the review was restricted to extrusion-based studies since this was the predominant additive manufacturing technique identified for hernia mesh development and to maintain a clearly defined and technically consistent scope. Beyond critically reviewing the available literature and compiling the processing parameters and properties evaluated in each study, this review aims to provide a practical resource for researchers working on the development of advanced hernia repair meshes.

2 Clinical context for hernia meshes

2.1 Hernia type and location

Hernias can be classified based on their anatomical location, etiology, size, and whether they are external or internal, with abdominal wall hernias being the most common in surgical practice. External hernias are characterized by a visible or palpable protrusion, while internal hernias occur when a portion of the bowel herniates through a defect in the peritoneal cavity, often presenting with few symptoms or bowel obstruction, which complicates

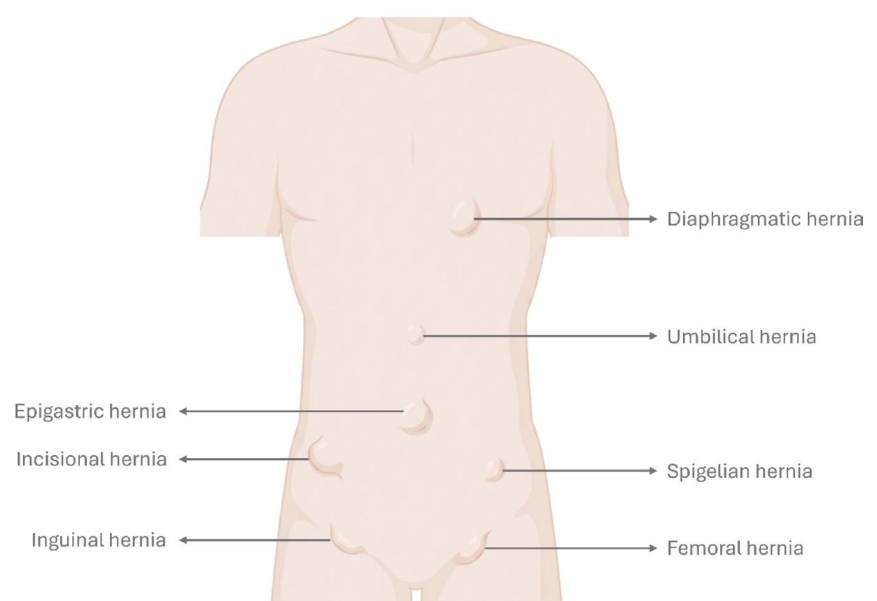
diagnosis [11, 12]. Anatomically, hernias can be divided into three main groups: groin hernias, ventral hernias, and diaphragmatic hernias (Fig. 1). Because these hernia types involve distinct anatomical structures and implantation environments, they impose different mechanical and biological requirements on the mesh.

Groin hernias occur in the inguinal region, which is the area where the anterior abdominal wall meets the thigh [6, 13]. They include direct and indirect inguinal hernias, which are differentiated by their point of entry into the inguinal canal (over the posterior wall of the inguinal canal or via the deep inguinal ring, respectively) and represent over two-thirds (70–75%) of all abdominal wall hernia procedures, and femoral hernias, which constitute nearly 6–17% of the cases [14–16]. These defects mainly involve the fascial and musculoaponeurotic structures of the groin, requiring meshes that provide sufficient reinforcement while remaining flexible and compliant to minimize discomfort and accommodate movement. Since the mesh is generally not placed in direct contact with the viscera, an anti-adhesive surface is typically unnecessary, and tissue integration and secure fixation are more relevant.

Ventral hernias arise in the anterior abdominal wall, excluding those in the inguinal region [6, 17]. The ventral hernia group is further subdivided into primary, such as epigastric, Spigelian, umbilical, and lumbar hernias, and secondary or incisional hernias [5, 16, 18], which typically develop at sites of previous surgical interventions due to incomplete tissue healing, infection, or excessive intra-abdominal pressure [17, 19], with incidence ranges from 11% to 20% after abdominal surgery [9, 17]. In terms of size, the European Hernia Society (EHS) framework classifies ventral hernias as small (< 2 cm), medium (between 2 and 4 cm), or large (> 4 cm), whereas incisional hernias require a separate width-based classification (W1 < 4 cm; W2: $\geq$ 4–10 cm; W3: $\geq$ 10 cm) [5]. These defects mainly involve the fascial and musculoaponeurotic structures of the abdominal wall and require a visceral anti-adhesion surface when intraperitoneal mesh placement is necessary.

Diaphragmatic hernias occur when abdominal structures protrude into the thoracic cavity through a defect in the diaphragm and are primarily classified as congenital or acquired [20]. Congenital diaphragmatic hernias result from incomplete fetal development of the diaphragm, with the most common subtype being the Bochdalek hernia [20]. Acquired diaphragmatic hernias develop later in life, most commonly following traumatic injury such as blunt or penetrating trauma, or as a complication of esophageal or gastric surgery [20]. Because these defects involve the diaphragm, meshes must be strong, flexible, and resistant to repeated respiratory motion. When the mesh is placed in contact with abdominal organs, an anti-adhesion surface may also be required.

Fig. 1 Different types of hernias



2.2 Hernia repair surgery

Surgical repair remains the only curative treatment for abdominal hernias, with more than 20 million procedures performed worldwide each year [5, 17]. Before the adoption of current treatment strategies, which rely on the use of an implant to reinforce the abdominal wall, hernia repair consisted exclusively of suture techniques, such as Bassini's method [5, 19]. These procedures aimed to close the defect by sewing the surrounding tissues together under tension [5, 14]. Although widely used, these techniques were associated with high recurrence rates and postoperative pain because of the excessive tension applied to the repaired tissue. These limitations created the need for new strategies, specifically the tension-free approach [5, 15], which can be achieved through three main surgical techniques: open repair, laparoscopic repair, and robotic-assisted repair. The selection between these techniques is influenced by the surgeon's expertise, the patient's condition, and the characteristics of the hernia [6, 21]. Other factors, such as the type of implant, technique-specific contraindications, and cost, also play critical roles in determining the most appropriate approach [21].

Open repair surgery can be performed with or without the placement of a prosthetic mesh, and it is commonly used for large hernias, particularly when reconstruction of the abdominal musculature or restoration of abdominal wall functionality is required. It allows mesh placement in different extraperitoneal planes and facilitates broad overlap. However, it is associated with greater tissue disruption and a higher risk of wound-related complications. Its application may also be limited when the rectus muscles show signs of atrophy or retraction, as these conditions hinder adequate mesh overlap, especially when the anatomical space has already been compromised by previous surgeries [21].

Laparoscopic repair surgery is particularly recommended for patients with small to moderately sized hernias and may be advantageous in patients at increased risk of wound-related complications. It is performed through small abdominal incisions using a laparoscope and specialized instruments. Postoperative complications associated with this technique are typically related to poor fixation, insufficient overlap, inadequate mesh properties, and patient-related factors such as obesity or persistent smoking. A significant limitation of this approach is that, in larger defects, the reconstructed abdominal wall may not regain adequate functionality. When the mesh is placed intraperitoneally, it should also provide a visceral anti-adhesion surface [21].

Robotic-assisted surgery has emerged as a minimally invasive alternative that applies principles similar to those of traditional laparoscopy. It has been used mainly for primary and uncomplicated incisional hernias and may facilitate suturing and abdominal wall reconstruction in selected cases. It is particularly suited for small to medium-sized hernias, especially in patients with complex abdominal anatomy or those requiring functional abdominal wall reconstruction. Recently, robotic techniques have been developed to release the transversus abdominis muscle (TAR), thereby expanding reconstruction possibilities. The main advantage lies in the articulated instrumentation, which allows for more precise suturing towards the abdominal wall, enabling the transformation of traditionally open procedures into less invasive interventions, with the potential to optimize clinical outcomes and accelerate recovery. However, it is generally associated with higher costs, longer operative times, and limited suitability for very large defects or patients with extensive surgical histories [21].

Overall, the selected surgical approach influences the required mesh dimensions, flexibility, surface properties, and fixation strategy. Additionally, although the use of surgical implants has significantly reduced hernia recurrence rates, complications such as infection and persistent pain still occur. Their incidence varies according to the type of hernia and surgical approach. Chronic postoperative pain after groin hernia repair has been reported in approximately 10 to 12% of patients, while debilitating pain affecting daily activities occurs in approximately 0.5 to 6% of cases [22]. To address these challenges, it is essential to consider implant properties and composition [2], highlighting the need for surgical implants that combine mechanical support with biocompatibility, flexibility, tissue integration, and multifunctionality.

3 Mesh structures for hernia treatment

As discussed in the previous section, the limitations of tension-based suture repair strategies were a key factor in the development of prosthetic reinforcement. A major clinical milestone in this transition occurred in 1958, when Dr. Francis Usher introduced the first successful synthetic polymer mesh, establishing the foundation for modern hernia treatment [6, 23]. A surgical mesh is a flexible implant placed at the defect site to cover and reinforce the weakened surrounding abdominal tissues. Its primary purpose is to significantly reduce hernia recurrence by providing the necessary support for tissue ingrowth and enabling tension-free closure.

Although advancements in surgical techniques, combined with the development of new advanced biomaterials, have transformed fibrous mesh into an essential element in hernia repair that promotes improved clinical outcomes, challenges remain [17]. Hernia recurrence remains a concern, with rates ranging from 5 to 10%, influenced by individual characteristics, surgical techniques, and the type of mesh used [2, 24]. A key biomechanical factor contributing to re-herniation is the compliance mismatch between many mesh materials and the highly deformable native abdominal wall [23].

Beyond recurrence, implants can cause other complications, such as foreign body response, which may lead to the formation of seromas or fibrosis, often accompanied by stiffness and chronic pain. In severe cases, these complications may require mesh removal and complete debridement of surrounding tissues. Furthermore, visceral tissue adhesions can be significant complications during wound healing involving surgical meshes [25]. In abdominal wall reconstruction, meshes with anti-adhesive properties can help prevent the formation of adhesions between the implant and abdominal organs, reducing the risk of bowel obstruction, chronic pain, and other complications [24]. Importantly, anti-adhesive meshes should limit visceral adhesion while preserving tissue integration on the parietal side [26].

To overcome these limitations, current mesh development focuses on surface modification, structural redesign, and changes in the materials used. These innovations aim to minimize deficiencies such as poor biocompatibility and introduce functional properties such as antimicrobial activity. Other studies in this field aim to improve drug loading and delivery at the surgical site to reduce the risk of postoperative infection. These modifications seek to enhance biological and mechanical properties and, consequently, mesh performance [24].

3.1 Mesh properties

Mesh structures used in the treatment of hernias are produced from fine, porous, and mechanically resistant materials [27]. Although an “ideal mesh” has not yet been developed, continuous advances in the field and the emergence of advanced materials provide the foundations for optimized design. Surgical meshes design should consider several interconnected requirements, such as mechanical strength and stability, appropriate pore architecture, tissue integration, ease of placement, and, when relevant, controlled degradation and surface bioactivity [26]. These properties must be balanced according to the type of hernia, anatomical site, implantation plane, and consequent mechanical requirements.

Darie-Niã et al. state that the general requirements necessary for the design of materials for medical applications are *in vitro* biocompatibility, mechanical properties, biodegradability, water absorption, and bioactivity [4]. Biocompatibility is essential because all implanted materials trigger a host response, which should remain controlled and clinically acceptable rather than causing excessive inflammation or adverse reactions. The use of innovative materials that adapt to physiological conditions contributes to improved biocompatibility and tissue integration [2]. To enhance the biocompatibility of structures, it is necessary to obtain a beneficial host response; cell-compatible materials such as silk fibroin (SF), gelatine, and hyaluronic acid may be used [25].

Biodegradability assumes significant importance in the development of structures for hernia repair [4]. At present, many of the structures used are not biodegradable, remaining indefinitely in the body or, in some cases, requiring a second surgical intervention for removal. This permanence may be associated with long-term complications such as chronic inflammation or rejection. Consequently, the possibility of designing a structure that

degrades in a controlled manner is highly relevant, provided that an effective balance can be achieved between the mechanical support required during the critical regeneration period and the subsequent degradation of the material, thereby avoiding the need for removal and minimizing the risks associated with the prolonged presence of foreign bodies in the organism.

As for mechanical strength, this is one of the fundamental characteristics of fibrous mesh-like structures, due to the maximum load applied to the abdominal wall after surgery. Based on empirically derived values reported in previous studies, the minimum required strengths for these structures are considered to be 22 N/cm and 32 N/cm in the longitudinal and transverse directions, respectively [24, 28, 29]. Most synthetic structures, regardless of weight and design, exhibit a minimum mechanical strength of 32 N/cm, although compliance with this reference value alone does not ensure adequate clinical performance [24]. The tensile strength of the structures must be sufficiently high to prevent rupture under mechanical loads in the human body. Under a load of 16 N, considered the typical force under physiological conditions, the intact abdominal wall extends 27% longitudinally and 16% transversely [25]. However, mesh performance also depends on compliance, anisotropy, resistance to cyclic loading and fatigue, creep, shrinkage, and fixation-related forces. It is essential that the structure provides sufficient strength and size to prevent retraction or displacement. Structural extensibility and flexibility help ensure compatibility with the abdominal wall. A structure that is either too rigid or too flexible may hinder the placement of the fibrous mesh, particularly in laparoscopic procedures [25].

The porosity of the material and the size of the pores influence bacterial growth and cell proliferation. Porosity affects tissue regeneration, as pores support tissue infiltration and integration with the mesh. Pores may be classified as very large ($> 2000\ \mu\text{m}$), large ($1000\text{--}2000\ \mu\text{m}$), medium ($600\text{--}1000\ \mu\text{m}$), small ($100\text{--}600\ \mu\text{m}$), or micropores ($< 100\ \mu\text{m}$), and may present diamond, hexagonal, or square geometries [5]. Small pores may restrict tissue infiltration and increase the inflammatory response, whereas the presence of large pores may result in a reduction of the contact area with the surrounding tissue, leading to increased pain during rest and physical activity [30]. Pore geometry may also influence tissue growth, with hexagonal pores reported to promote greater infiltration than square or diamond-shaped pores [24].

The weight of the structures is also an important aspect: meshes may be classified as heavyweight ($> 90\ \text{g/m}^2$), medium-weight ($50\text{--}90\ \text{g/m}^2$), lightweight ($35\text{--}50\ \text{g/m}^2$), or ultralight ($< 35\ \text{g/m}^2$) [5]. Lightweight meshes contain less implanted material and may therefore reduce the foreign-body response. However, mesh performance depends on the combined effects of weight, pore architecture, mechanical stability, fixation, and defect characteristics [24].

In addition to responding to mechanical properties, mesh structures should be easy to handle, withstand the sterilization procedure, and demonstrate low toxicity (or preferably non-toxic behavior) at the cellular level [4]. Clinical risks such as recurrence, pain, and infection should be assessed through a combination of mechanical testing, biological evaluation, animal models, sterilization validation, and clinical follow-up [31]. The production technique also influences the properties of the mesh. The most common design of these devices is the mesh structure, with most meshes manufactured through the knitting process. Alternative techniques, such as advanced manufacturing techniques, including electrospinning and FDM-based 3D printing, can provide greater control over mesh architecture, surface characteristics, and functional performance, supporting the development of more customized structures for hernia repair [32].

3.2 Type of meshes

Nowadays, meshes for hernia repair are commonly classified into distinct groups, which distinguish themselves from each other based on their material origin, mechanical properties, and biological interaction with the host tissue [14, 33]. While the literature offers various classification systems, in this review, a division into four types will be made: synthetic non-absorbable, synthetic absorbable, biological meshes, and composite and functionally modified meshes.

3.2.1 Synthetic non-absorbable structures

Synthetic meshes are the most commonly used prosthetic materials in hernia repair due to their excellent tensile strength, elasticity, and cost-effectiveness [6, 19]. These meshes can be subdivided into non-degradable (permanent) and degradable (absorbable) categories. Non-absorbable synthetic meshes, also referred to as traditional or classic meshes, provide permanent structural reinforcement, as they remain in the body indefinitely [18, 24]. The most widely used materials in this class include polypropylene (PP), polyethylene terephthalate (PET), expanded polytetrafluoroethylene (ePTFE), polyvinylidene fluoride (PVDF), polyurethane (PU), nylon-6 (PA6), and cross-linked acrylic polymers [6, 19, 23]. Table 1 presents some commercial structures and their composition.

PP is frequently used because of its flexibility and ability to promote good tissue integration, although it may induce chronic inflammatory reactions in some cases. ePTFE, on the other hand, is valued for its low adhesiveness and is often used in areas in contact with viscera, despite its lower porosity, which limits cellular incorporation. PET offers good mechanical properties but may be associated with higher rates of infection and inflammation [15].

Table 1 Synthetic non-absorbable commercial structures

Trade name	Composition
3DMax	PP
Atrium	PP
Bard Mesh	PP
Bard Teflon	PTFE
BardSoft	PP
Dacron	PET
Dipromesh	PP
DualMesh	e-PTFE
DualMesh Plus	e-PTFE
Dulex	e-PTFE
DynaMesh	PVDF
Fluoropassiv	PET
Gore-tex	e-PTFE
Infinet Mesh	PTFE
Intramesh T1	e-PTFE
Lars	PET
Marlex	PP
Mersilene	PET
MicroVal 2D Mesh	PP
MotifMESH	e-PTFE
MycroMesh	e-PTFE
Optilene	PP
Parietene	PP
Parietene Light	PP
Parietex	PET
PerFix plug	PP
Polysoft	PP
Prolene	PP
Prolite	PP
Serapren	PP
Surgimesh	PP
Surgipro	PP
TecnoMesh	PP
Trelex	PP
Visilex	PP
Vitamesh	PP

Despite their advantages, including stable mechanical properties, high tensile strength, and durability, non-absorbable meshes are associated with long-term complications. These include chronic pain, persistent inflammation, infection, visceral adhesions, stiffness, and enterocutaneous fistula formation [16]. The selection of non-absorbable materials is typically based on the need for permanent structural support, particularly in clinical contexts where the abdominal wall shows significant fragility or a high risk of recurrence.

3.2.2 Synthetic absorbable structures

Absorbable synthetic meshes were developed to mitigate some of the complications associated with non-absorbable implants. These meshes are designed to provide temporary mechanical support during the healing period, gradually degrading once their function has been fulfilled [6, 19]. They are characterized by their ability to degrade over time, generally through hydrolysis or enzymatic action, under physiological conditions such as body temperature, the presence of biological fluids, and neutral pH. However, their degradation profiles vary considerably, ranging from rapidly absorbable materials that lose mechanical support within a relatively short period to slowly absorbable or biosynthetic polymers designed to maintain reinforcement for several months.

Common absorbable synthetic polymers include polyglycolic acid (PGA), polylactic acid (PLA), poly(lactic-co-glycolic acid) (PLGA), polycaprolactone (PCL), poly-4-hydroxybutyrate (P4HB), polydioxanone (PDO), poly (ester urethanes) (PEUs), and poly(trimethylene carbonate) (PTMC) [16, 19]. These materials may be derived from natural monomers such as PLA or from petrochemical-derived monomers via chemical synthesis, such as PCL [34, 35]. Table 2 presents some commercial absorbable structures.

Although absorbable meshes offer the advantage of reducing long-term foreign body reactions, their degradation over time presents some challenges. The mesh may lose mechanical strength before the tissue has regenerated sufficiently to prevent recurrence [5, 6]. Therefore, precise control over the degradation rate is essential to ensure that these materials retain adequate mechanical integrity during tissue regeneration, highlighting the importance of carefully tailoring degradation kinetics for clinical success.

Currently, absorbable synthetic polymers still have limited commercial use, not only due to their higher cost, but also because they generally do not achieve mechanical strength levels comparable to non-absorbable materials. However, combined structures that integrate degradable and non-degradable polymers are commercially available to optimize clinical performance and will be discussed in the following sections. On the other hand, the use of absorbable synthetic materials is already quite common in auxiliary functional components used in hernia treatment. PLA is employed in resorbable monofilament pins or hooks, while PLGA and PDO are used in expandable and retraction rings, respectively. PET is used in self-expanding rings, and bioabsorbable polyacrylates may be applied as coatings to provide additional rigidity [31].

3.2.3 Biological structures

Biological meshes are made from decellularized ECM, which can be obtained from human (allogenic/allograft) or animal (xenogeneic/xenograft) tissues [23, 36]. The most common sources include human dermis, porcine

Table 2 Commercial absorbable structures

Trade name	Composition
Dexon	PGA
DuraSorb	PDO
CoSeal	PEG
Guidor	PLA
NeoveilT	PGA
Phasix	P4HB
Poly lactide Mesh	PLA
Sefil	PGA
Vicryl	PLGA

dermis or small intestinal submucosa, bovine dermis or pericardium, and equine pericardium [5, 18]. Table 3 presents some commercial biological structures.

These meshes serve as bioactive scaffolds that promote wound healing and regeneration through cellular repopulation and neovascularization, facilitated by the infiltration of native fibroblasts and the formation of new connective tissue [6, 36]. Because of their biocompatibility and lower inflammatory response, these meshes are normally proposed for complex, contaminated, or infected surgical fields [16, 18].

Notwithstanding their advantages, these meshes have notable limitations, including high cost, limited suitability for large-scale production, lower mechanical strength, and susceptibility to stretching over time due to elastin (ES) retention, which can lead to hernia recurrence [15, 37]. Additionally, studies report significant differences between biological meshes. Human-derived meshes, for example, tend to have higher recurrence rates than xenograft alternatives, and cross-linked porcine dermal meshes exhibit increased inflammatory responses compared with non-cross-linked ones [18].

3.2.4 Composite and functionally modified meshes

Composite meshes combine two or more different materials or layers with complementary functions. Functionally modified meshes may overlap with this category and are based on existing structures that acquire additional properties through coatings, chemical treatments, or specific architectural features. Examples of commercially available composite meshes are shown in Table 4.

Most composite meshes are based on a mechanically robust synthetic layer, commonly composed of PP, PET, PVDF, or ePTFE, combined with an absorbable polymer or biological matrix. This combination can overcome some of the limitations of the individual components and enable different regions or layers of the implant to perform complementary functions. For instance, these designs can incorporate self-gripping components, antimicrobial agents, or other additives. Increasing research has focused on naturally derived biopolymers, such as polyhydroxyalkanoates (PHA), chitosan (CS), alginates, and protein- or polysaccharide-based materials such as silk fibroin, collagen, hyaluronic acid, cellulose, elastin, gelatin, and keratin [26]. These materials have attracted increasing interest due to their biodegradability, biocompatibility, intrinsic biological properties, and capacity to

Table 3 Commercial biological structures

Trade name	Composition
AlloDerm	Human dermis
AlloMax	Human dermis
Cellis	Porcine dermis
CollaMend	Porcine dermis
Cortiva	Human dermis
Epiflex	Human dermis
Flex HD	Human dermis
FortaGen	Porcine intestine
Fortiva	Porcine dermis
Gentrix surgical matrix	Porcine Urinary bladder
Miromesh	Porcine Liver Tissue
Peri-Guard	Bovine Pericardium
Permacol	Porcine Dermis
Strattice	Porcine Dermis
SurgiMend	Bovine Fetal Dermis
Surgisis	Porcine intestine
Surgisis/Biodesign	Small Intestinal Submucosa
Tutopas	Bovine pericardium
Tutopatch	Bovine Pericardium
Veritas	Bovine pericardium
XCM Biologic	Porcine dermis
Xenmatrix	Porcine Dermis

Table 4 Commercial composite structures

	Trade name	Composition
	C-Qur	PP / omega-3
	Composix EX / Dulex	PP / e-PTFE
	DynaMesh-IPOM	PP / PVDF
	GORE BIO-A	PGA / PTMC
	OviTex	PP / Ovine ECM reinforcement
	Parietene DS	PP / collagen
	Parietex Composite	PET / collagen
	Phasix ST	P4HB / HA / CMC / PEG
	Physiomesb	PP / PGC-25 / PDO
	Proceed	PP / ORC / PDO
	ProGrip	PET / PLA
	Sepramesb	PP / HA / CMC
	Symbotex	PET / collagen
	Synecor	PP / PTFE / PGA
	Teramesb Composite	PP / PGA
	Ti-Mesh	PP / Ti
	TIGR	PLA / PTMC
	Ultramesb Composite	PET / PE
	Ultrapro	PP / PGC-25
	VentraleX	PP / e-PTFE
	Ventralight ST	PP / HA / CMC / PEG / PGA
	Ventrio	PP / e-PTFE
	Vicryl (Composite)	PP / PG 910
	Vypro	PP / PG 910
	Zenith Composite	PET / PU

HA: sodium hyaluronate; CMC: carboxymethylcellulose; PEG: polyethylene glycol; PGC-25: poliglecaprone 25; PDO: polydioxanone; ORC: oxidized regenerated cellulose; PTFE: polytetrafluoroethylene; Ti: titanium; PG 910: polyglactin 910

promote tissue regeneration through enhanced cell adhesion, proliferation, and extracellular matrix deposition [24, 26]. However, their limited mechanical strength and stability often restrict their use as standalone meshes. Their integration as coatings or membrane layers supported by a mechanically robust synthetic structure may therefore provide a more viable strategy for preserving biological functionality while maintaining the mechanical performance required for abdominal wall repair.

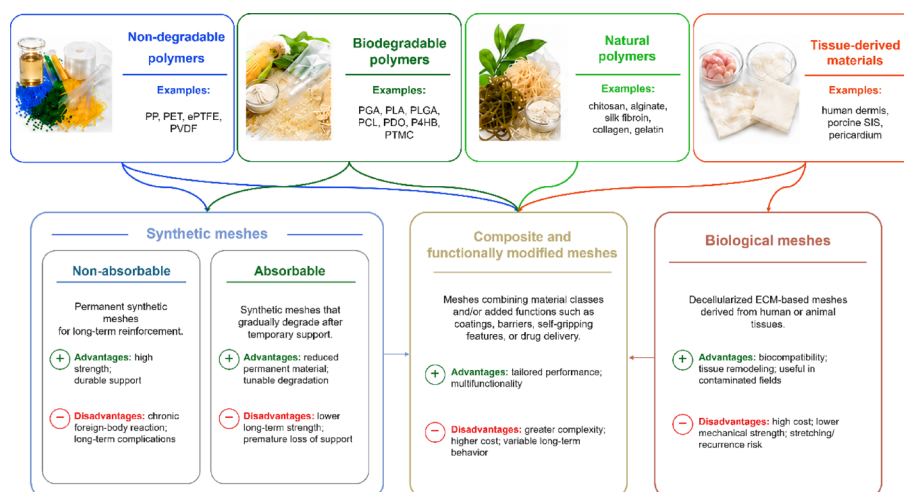
Composite and functionally modified meshes represent a promising strategy for the development of next-generation implants combining mechanical support with biological and therapeutic functions. Additive manufacturing may enhance this approach by enabling precise control over material distribution and mesh architecture, although challenges related to interlayer adhesion and degradation rates of the individual components must still be addressed.

Figure 2 summarizes the main advantages and limitations of the different mesh groups, as well as the materials that can be used in each type of implant. Non-absorbable meshes are the most commonly used and provide mechanical support but are associated with chronic inflammation and adhesion formation. Absorbable meshes may reduce the long-term foreign body burden, although premature degradation may compromise sustained reinforcement. Biological meshes promote tissue regeneration and may be suitable for contaminated fields, but their lower durability and tendency to creep remain important limitations. Composite meshes aim to balance these properties by combining structural support with biological or functional components, although their increased complexity may introduce challenges related to interlayer compatibility and higher manufacturing cost.

3.3 Mesh fixation

Mesh fixation technique is considered one of the main challenges of hernia repair. Fixation may be performed using sutures or staples, which involve tissue perforation and may lead to inflammation, injury, and chronic pain.

Fig. 2 Classification of hernia mesh materials and their integration into synthetic, biological, and composite or functionally modified meshes, highlighting the main advantages and limitations of each group



Inadequate fixation can result in mesh migration and subsequent hernia recurrence [15, 27]. There are two types of sutures: absorbable and non-absorbable. Absorbable sutures lose strength too quickly, potentially causing hernia recurrence. Non-absorbable sutures retain strength indefinitely but are associated with increased chronic pain and longer recovery times. However, non-absorbable sutures show lower recurrence rates in laparoscopic surgeries, as they maintain mesh integrity until the abdominal wall is fully healed [15].

In addition to sutures, meshes can be fixed using fibrin-based glue or cyanoacrylate glue. The two main components of fibrin glue are fibrinogen and thrombin. Fibrinogen may be autologous (from the patient) or allogeneic (from a donor). Fibrin glue can be enhanced with substances such as growth factors, platelets, and antibiotics. Fibrinogen provides tensile strength and adhesive properties, with its original function being hemostatic, helping to stop bleeding by promoting blood clotting. Unlike fibrin glue, cyanoacrylate glue is synthetic and classified as a hybrid tissue sealant. It offers rapid adhesion and strong bonding to biological tissues. Currently, there is no clear evidence indicating which glue performs better for mesh fixation. There is no significant difference in chronic pain or hernia recurrence between cyanoacrylate glue and absorbable sutures in the long term. However, early chronic pain is less frequent with glue fixation than with sutures. Glue fixation is also considered faster and less painful in open surgeries. Although various materials are available for mesh fixation, there are no clinically significant differences between them [15]. Table 5 lists some of the fixation methods and the materials used.

As previously mentioned, the fixation technique for mesh structures remains a major challenge in surgical procedures. To address this issue, self-adhesive meshes are being developed for hernia repair, as they reduce the potential for persistent postoperative pain. These self-adhesive meshes aim to minimize both acute and chronic postoperative discomfort, since they eliminate the need for sutures or other fixatives, which are often associated with chronic pain. Consequently, patients treated with these meshes are expected to resume daily activities sooner and with reduced reliance on postoperative analgesics [38].

An example of a self-adhesive mesh is described by Martinez et al., who present Adhesix® as a partially resorbable and hydrophilic structure designed to cover extraperitoneal defects of the abdominal wall. The non-resorbable portion of the mesh consists of 60% monofilament poly-L-lactic acid (PLLA) and 40% monofilament PP. The underside of the mesh contains a bioadhesive gel that adheres to the abdominal wall upon contact with moist tissue. This gel is composed of polyvinylpyrrolidone (PVP) and PEG [39].

The use of self-adhesive meshes raises a relevant clinical challenge. While self-adhesion enables effective fixation without the need for sutures, many studies report that anti-adhesive properties are particularly beneficial, especially in reducing complications such as fibrosis and visceral adhesion. Therefore, a compromise between a self-adhesive layer in direct contact with the regenerating tissue and an opposite face with anti-adhesive characteristics, designed to prevent adhesion to the other abdominal tissues such as the viscera, may represent a promising

Table 5 Fixation methods and materials. (Adapted from [15]).

Fixation method	Type of fixation	Product (manufacturer)	Material
Tissue adhesive	Fibrin glue	Tisseel (Baxter)	Human fibrinogen, synthetic aprotinin, human albumin, <i>L</i> -histidine, niacinamide, polysorbate 80, disodium chloride dihydrate, human thrombin, calcium chloride dihydrate, sodium chloride, water for injection
Tissue adhesive	Fibrin glue	Evicel (RxList)	Human fibrinogen, human thrombin, calcium chloride, sodium chloride water for injection
Tissue adhesive	Cyanoacrylate glue	Histocryl (B. Braun)	<i>n</i> -butyl-2-cyanoacrylate monomer
Tissue adhesive	Cyanoacrylate glue	Glubran-2 (GEM Srl)	<i>n</i> -butyl-2-cyanoacrylate and formaldehyde monomer from GEM Srl
Sutures	Absorbable sutures	Monocryl (J&J)	Polyglecaprone
Sutures	Absorbable sutures	Vicryl (J&J)	Polyglactin
Sutures	Absorbable sutures	Dexon (N.A)	Polyglycolic acid
Sutures	Absorbable sutures	Maxon (Medtronic)	Polyglyconate
Sutures	Absorbable sutures	PDS (J&J)	Polydioxanone
Sutures	Non-absorbable sutures	Prolene (J&J)	Polypropylene
Sutures	Non-absorbable sutures	Nylon (Dolphin Sutures)	Polyamide

Table 6 Clinical requirements for hernia mesh design and manufacture

Clinical scenario	Main clinical challenges	Adhesion requirements	Key design requirements	Potential manufacturing advantages
Groin hernia	Dynamic loading, fixation	Controlled tissue integration	High flexibility and compliance; low stiffness; controlled anisotropy	3D printing allows control of structure and pore geometry; electrospinning allows tissue integration
Ventral/incisional hernia	Multidirectional loading, risk of recurrence	Tissue integration on the parietal side; visceral anti-adhesion surface	High load-bearing capacity; multidirectional strength; fatigue resistance	3D printing allows structural control and anti-adhesion layer; hybrid systems allow differentiated layer functions
Diaphragmatic hernia	Continuous cyclic motion, complex anatomy	Controlled tissue integration; visceral anti-adhesion surface when necessary	High flexibility; cyclic fatigue resistance; geometrical adaptability	3D printing enables controlled architecture; hybrid systems allow differentiated layer functions

design strategy for this type of mesh. However, the required balance between fixation and anti-adhesion must be adapted to the specific hernia type and anatomical location, as discussed in the previous sections. Advanced manufacturing techniques may support this approach by enabling the adhesive and anti-adhesive multilayer structure.

3.4 Clinical requirements for mesh design and manufacture

The clinical requirements of hernia repair depend on the hernia type, anatomical location, and expected interaction between the mesh and surrounding tissues, as discussed on the previous sections. These factors directly influence the required mechanical, biological, and fixation properties of the implant and, consequently, the potential role of different manufacturing strategies. Accordingly, mesh design should be tailored not only to provide adequate mechanical reinforcement but also to address the biological and functional demands associated with each specific clinical scenario. Table 6 summarizes the relationship between representative clinical scenarios, the corresponding mesh design requirements, potential manufacturing approaches, and the most relevant evaluation endpoints.

4 3D printing in the production of fibrous mesh structures for hernia treatment

4.1 Fundamentals of 3D printing

Since its emergence in 1986, 3D printing technology has been developed over the years, giving rise to a wide range of methods and techniques [19]. 3D printing enables the production of a device through the layer-by-layer deposition of material. In this process, a consecutive series of two-dimensional images is created using computer-aided design (CAD) software, which is then used to generate a digital model that produces a 3D object with precise geometry and shape. According to the American Society for Testing and Materials (ASTM) and the International Organization for Standardization (ISO), current 3D printing strategies are classified into seven procedures: (i) Material Extrusion (including FDM); (ii) Powder Bed Fusion; (iii) Vat Photopolymerization; (iv) Material Jetting; (v) Binder Jetting; (vi) Sheet Lamination; and (vii) Directed Energy Deposition [19]. During the production process, certain parameters may be adjusted, such as wall thickness, extrusion temperature, printing speed, and infill percentage [40].

The choice of process depends on factors such as the type of device to be produced, its application, manufacturing materials, production time, cost, equipment availability, and technical expertise [19]. FDM is the most widely used technique, as it offers greater simplicity, compatibility with various materials, and lower cost. In FDM, a thermoplastic filament passes through a heated print head, where it is extruded and deposited onto the build platform, serving as the base for the formation of successive layers until the object acquires its final shape. FDM has the capacity to construct structures rapidly, with dimensional accuracy and suitable mechanical properties. These characteristics make this technique widely used in the manufacture of personalized medical devices tailored to each patient [41].

Nevertheless, other 3D printing technologies may provide specific advantages, particularly in terms of resolution, surface finish, and the ability to process different classes of materials. Vat photopolymerization techniques, including stereolithography (SLA), digital light processing (DLP), and liquid crystal display (LCD) printing, enable the fabrication of structures with higher resolution and smoother surface finishes than the structures typically obtained by FDM [42]. However, they require extensive post-processing, and their use in biomedical applications may be limited by the restricted availability of suitable biocompatible resins and potential cytotoxicity [43, 44]. Selective laser sintering (SLS) enables the fabrication of complex structures without support materials but is limited by high equipment costs and a restricted range of suitable polymer powders [45].

In the specific case of biomedical applications, extrusion-based bioprinting enables the fabrication of complex structures based on natural, biological, and biocompatible materials that cannot be processed by high-temperature thermoplastic extrusion due to the temperatures involved. This low-temperature extrusion approach may support tissue regeneration by reproducing ECM-like environments, while also enabling the fabrication of cell-laden structures through the direct incorporation of living cells into the bioink [46, 47]. Nevertheless, its application remains limited by the low mechanical strength of the structures and the lack of long-term structural stability [48]. For cell-incorporated structures, although this technique enhances regenerative potential, maintaining cell viability during processing remains a significant challenge [47]. For hernia repair, additional challenges include ensuring that the cell-containing structure can withstand the dynamic loads of the abdominal wall without compromising cell viability or regenerative function.

At present, many medical devices are being produced using 3D printing, as its customized design allows the fabrication of complex anatomical structures of the human body, as well as devices adapted to the needs of each patient. In biomedical applications, extrusion-based 3D printing is the most widely used technique [19]. This technology has become attractive in the biomedical field due to its cost-effectiveness and the possibility of obtaining personalized devices within relatively short timeframes [8]. Thus, it is possible to produce customized implants for hernia repair, potentially improving anatomical conformity. It also enables the creation of anatomical models for pre-operative planning, thereby increasing surgical precision [2]. Although other 3D printing

technologies offer specific advantages in terms of resolution, surface finish, and the processing of different materials, the literature search was restricted to extrusion to keep a consistent scope and because it is the predominant approach reported for hernia mesh development.

4.2 Fibrous mesh structures produced by 3D printing

3D printing, specifically FDM, has emerged as a groundbreaking technology for developing medical devices for hernia repair, due to its precision and high customization. However, a critical analysis of recent literature shows that the successful clinical application of 3D-printed meshes requires a complex balance between bio-functionalization, structural mechanics, and biological integration. Table 7 lists the reported cases of fibrous mesh structures produced via 3D printing.

Traditional polymers used in medical implants, such as polyvinyl alcohol (PVA), PCL, PP, and PLA, are synthetic polymers with limited intrinsic bioactivity and susceptible to multiple postoperative complications. To address this problem, both Ballard et al. and Qamar et al. demonstrated the feasibility of embedding broad-spectrum antibiotics directly into thermoplastic filaments prior to printing [40, 49]. Ballard et al. incorporated gentamicin into PLA filaments, and through agar diffusion assays, they confirmed that the printed meshes produced zones of inhibition against *Staphylococcus aureus* and *Escherichia coli*, validating the presence of the antibiotic despite the high temperatures used in the printing process [49]. However, as their study was limited to in vitro characterization, subsequent in vivo assays are necessary because drug release and antimicrobial activity can vary due to physiological variables, such as cellular interactions, dynamic fluid, and protein absorption. Qamar et al. advanced beyond this limitation by incorporating ciprofloxacin (CIP) into both PVA and PP filaments and validating their approach in a rabbit abdominal wall defect model [40]. The results suggest that drug-loaded printed meshes could reduce inflammation and accelerate healing in living tissue, while thermal analysis confirmed that the printing process did not degrade the antibiotic [40]. Nevertheless, the rapid initial drug release of 60–70% observed within the first hour of implantation raises questions regarding the impregnation method, as it may fail to maintain the therapeutic concentration required to prevent late-onset infections.

To overcome this limitation of simple impregnation, Calero Castro et al. investigated the encapsulation of gentamicin in an alginate hydrogel as a strategy for achieving a more sustained drug release profile from PCL meshes [50]. The results show that alginate encapsulation forms uniform inhibition halos against *Escherichia coli*, contrasting with asymmetric zones generated by unencapsulated antibiotic (Fig. 3), demonstrating that hydrogel barriers could modulate drug diffusion. In vivo results indicated only mild acute and chronic inflammatory responses with substantial fibroblastic migration, collagen deposition, and neovascularization. However, this study presents a critical limitation, the absence of drug release assays to quantify gentamicin kinetics, and the seven-day implantation period provides no information about long-term tissue integration or chronic host response.

Olmos-Juste et al. used a similar strategy to Calero Castro et al. They developed an ink that could be extruded at ambient temperature, made of waterborne polyurethane (WBPU), sodium alginate, and chloramphenicol [51]. Their results indicate a release of 80% of the antibiotic within the first twenty-four hours, which, when compared with the results obtained by Qamar et al., shows a significant improvement in release kinetics, demonstrating that alginate encapsulation effectively prolongs drug elution. Nonetheless, a critical limitation of this study, as in the Ballard et al. study, is that the assays were only performed in vitro. As mentioned before, physiological conditions can influence essential aspects of drug release and mesh performance, so in vivo evaluation is essential. In addition to drug release, their formulation was also designed to improve mechanical performance [51]. The printed meshes exhibited isotropic behavior, while the authors argue that this is a clinical advantage since it facilitates surgical fixation in any direction, it fails to mimic the native abdominal wall, which presents anisotropic behavior. Nonetheless, the group achieved physiological elongation and yield limits close to those observed in the human abdominal wall.

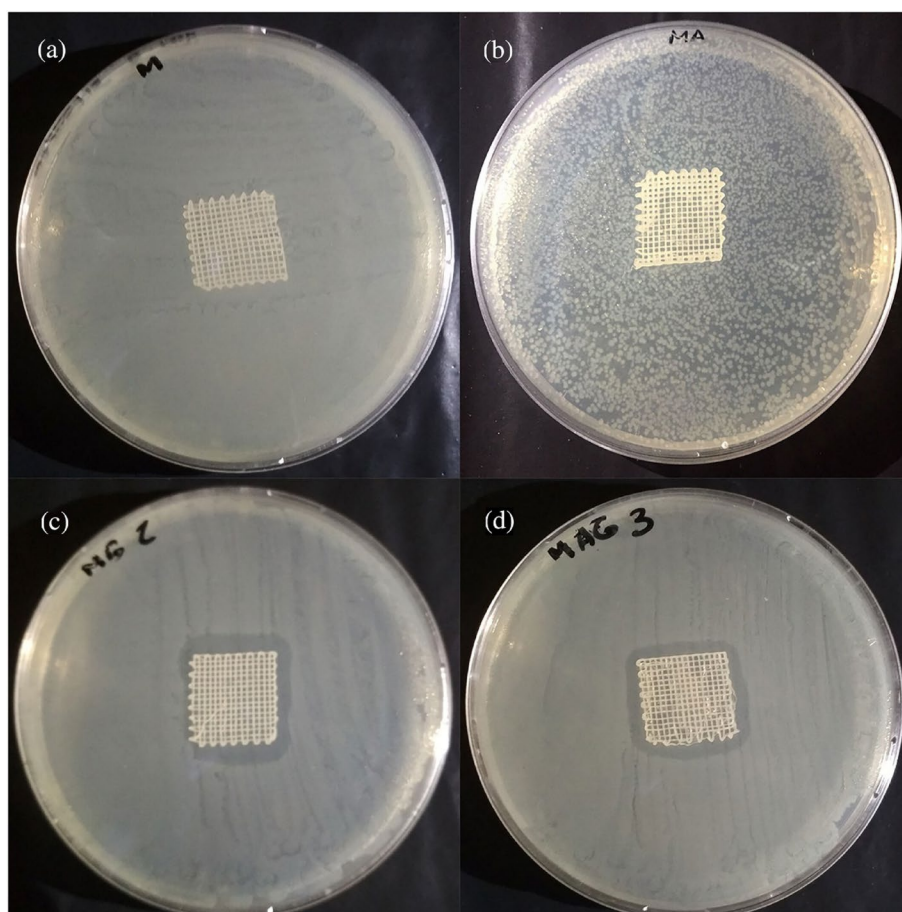
Table 7 Overview of the mesh structures produced by 3D printing, including the materials used, structure and dimensions, processing parameters, and evaluated functional properties

Reference	Materials	Structure and dimensions	Parameters	Functional properties evaluated
[40]	PP PVA CIP	Dimensions: 100 × 100 mm Width: 0.8 mm and 1.2 mm Pore size: 2 mm and 3 mm Pore shape: Square and diamond	PP: Temperature: 190 °C Speed: 90 mm/s Layer height: 0.2 mm Infill: 90% PVA: Temperature: 200 °C Speed: 150 mm/s Layer height: 0.2 mm Infill: 90%	Drug loading and release characteristics Mechanical properties Thermal analysis In vivo Biocompatibility, adhesiogenicity and histologic characterization
[49]	PLA Gentamicin	Large pores	-	Morphological evaluation with Scanning Electron Microscopy (SEM) Bioactive properties Bacterial inhibition
[50]	PCL SA Gentamicin	Dimensions: 2 × 2 cm Pore size: Micropores – 0.75 × 0.75 mm; Macropores – 1.25 × 1.25 mm;	PCL: Speed: 10 mm/s Flow rate: 1 mm/s SA: Speed: 10 mm/s Flow rate: 5 mm/s	Bacterial properties Histological behavior Adhesion
[51]	WBPU SA Chloramphenicol	Dimensions: 50 × 58 × 0.65 mm Pore size: 2.8 mm Pore shape: hexagonal	Print speed: 8 mm/s Nozzle/needle diameter: 0.8 mm	Morphological characterization Rheological characterization Mechanical characterization In vitro degradation Cell viability Swelling ratio In vitro drug release
[52]	PP	Pore size: 0.08 and 0.2 mm ² Pore shapes: rectangular; triangular, hexagonal Four and seven layers	Nozzle Temperature: 215 °C Bed Temperature: 90 °C Speed: 45 mm/s (infill), 28 mm/s (perimeters) Layer height: 80 µm	Mechanical properties Density Pore size and morphological characterization
[53]	PU	Pore diameters: 0.1–1 mm Pore shape: hexagonal Thickness: 0.5 mm Surface area: 8 × 12 cm ² Weight: 20–30 g/m ²	Temperature: 200–230 °C Speed: 30–120 mm/s Layer height: 0.1–0.3 mm	Morphological evaluation with SEM Mechanical properties Cell viability/Cytotoxicity Surface quality Sterilization efficacy
[54]	PLA TPU ADM	Dimensions: 26 × 26 × 6 mm Pore sizes: 2, 2.5, 3, 3.5, 4 mm Pore shapes: circular, diamond, square	-	Morphology and Topology Structural analysis with FTIR Mechanical properties Wettability measurement In vitro biocompatibility In vivo performance
[55]	PCL	Dimensions: 1 × 45 × 45 mm Pore size: 1 to 100 µm	Nozzle Temperature: 220 °C Bed Temperature: 60 °C Print speed: 8 mm/s	Morphology and porosity evaluation with SEM Wettability measurement Mechanical properties Biodegradation rate Cellular viability and proliferation In vivo biocompatibility Structural analysis with FTIR
[56]	Gelatin type B MWCNTs	Dimensions: 35 × 35 × 2 mm Pore size: 1 mm Four layers	Temperature: 25 °C Layer height: 0.5 mm Extrusion pressure: 55–110 kPa Print speed: 8 mm/s Infill density: 20%	Morphology and microstructure analysis with SEM Swelling behavior Hydrolytic degradation Wettability measurement Structural analysis with FTIR Rheological characterization Mechanical properties In vitro biocompatibility

Table 7 (continued)

Reference	Materials	Structure and dimensions	Parameters	Functional properties evaluated
[58]	PCL Barium Iodinated contrast Gadolinium-based contrast	Dimensions: 2×2 cm Large pores	Nozzle temperature: 125–130 °C Speed: 7 mm/s Layer height: 0.2 mm	Radio-opacity and stability with Computed tomography

Fig. 3 Antibacterial activity of gentamicin-loaded 3D-printed PCL meshes assessed by agar diffusion. **(a)** PCL mesh. **(b)** PCL mesh with alginate and calcium chloride. **(c)** PCL mesh and gentamicin. **(d)** PCL mesh with alginate, gentamicin, and calcium chloride. Reproduced with permission from reference [50]



Other studies have focused primarily on optimizing the mechanical performance of printed meshes, a critical parameter given the dynamic and repetitive stress of the abdominal wall. Galván-Chacón et al. compared a printed PP mesh with a commercial knitted PP mesh, finding fundamental limitations of FDM [52]. The results showed that although the FDM allows precise control over pore architecture, meshes made by this method exhibited significantly inferior mechanical properties. This can be due to the fixed junction points where orthogonal layers intersect, which prevent the multidirectional load distribution characteristic of knitted commercial meshes.

Wang et al. focused on defining an optimal pore geometry and print parameters for PU meshes [53]. The results indicated that with a layer height of 0.1 mm, a velocity of 60 mm/s, and a temperature between 210 and 220 °C, it is possible to produce implants with precise dimensional accuracy. Hu et al. developed a hybrid mesh combining PLA, thermoplastic polyurethane (TPU), and acellular dermal matrix (ADM), aiming to develop a structure with improved elasticity and biocompatibility [54]. The results indicated that TPU improved elasticity, while acellular dermal matrix enhanced biocompatibility by favoring cell proliferation and tissue growth. They also found that the diamond-shaped pore geometry with 3 mm pores provided optimal mechanical performance. In vivo evaluation demonstrated that commercial PP meshes have a higher adhesion score than the mesh developed

by them. It is important to note that this study presents some limitations. The use of umbilical vein endothelial cells (HUVECs) is not representative of the fibroblasts of the abdominal wall. As in the previous article, this mesh design also failed to mimic the anisotropic behavior of the native abdominal wall.

Navaei et al. tried to overcome the limitations of both biological and synthetic meshes by combining a synthetic material, PCL, and a biological one, gelatin [55]. This hybrid approach used FDM for the structural part and freeze-drying for the porous matrix. The results demonstrated that PCL provided the necessary structural integrity, while the gelatin promoted fibroblast attachment and proliferation. However, it should be noted that the meshes were placed subcutaneously rather than in the diaphragm, thereby not replicating the respiratory stress, dynamic multidirectional forces, and specific microenvironment of the native diaphragm. Yadav et al. also tried to overcome the limitations of both synthetic and biological meshes by developing an ink made of carboxyl-multi-walled carbon nanotube (MWCNTs) and gelatin [56]. The results showed that the incorporation of MWCNTs generally increased the ultimate tensile strength and elastic modulus while reducing the swelling ratio and slowing hydrolytic degradation over six weeks (Fig. 4). However, the mechanical improvement was concentration-dependent, and the differences between some formulations were not statistically significant, preventing a direct dose–response relationship from being established. The developed meshes were also reported to promote *in vitro* cell attachment and viability, although the use of MWCNTs introduces significant concerns regarding long-term biosafety and potential cytotoxicity [57]. Besides, the seven-day assessment provides no information about chronic inflammatory responses or degradation products.

Beyond mechanical and biological challenges, traditional hernia meshes are largely invisible on postoperative scans. Regarding this limitation, Ballard et al. used 3D printing to directly incorporate contrast agents (iodinated, gadolinium, and barium) into PCL meshes [58]. The results demonstrated that the three contrast agents showed initial radiopacity. However, barium was the most stable and effective agent for maintaining short-term computed tomography (CT) visibility, over the evaluated period (7 days), at human body temperature (Fig. 5).

Overall, the analyzed studies demonstrate that 3D printing, particularly FDM, offers a versatile platform for developing customized hernia meshes with precise control over mesh architecture and the capacity for multifunctionalization. Most of the reviewed articles try to overcome the mechanical limitations inherent to printed structures, which Galván-Chacón et al. identified as fixed junction points that prevent the multidirectional load distribution characteristic of knitted commercial meshes [52]. Researchers have tried different approaches, such as blending rigid polymers like PLA with elastomers like TPU or reinforcing soft biological matrices with 3D-printed synthetic skeletons, resulting in meshes with high elasticity and tensile strength.

In the studies reviewed, extrusion-based mesh fabrication predominantly relies on conventional thermoplastics, including PLA, PCL, PP, or PU. These polymers are easier to process because they are melted, deposited, and solidified by cooling, without generally requiring post-printing crosslinking. By contrast, the extrusion of bio-material-based formulations commonly requires low-temperature processing to preserve biological functionality.

Fig. 4 Results of mechanical testing showing (A) Ultimate strength and (B) Modulus of samples A (0 g MWCNTs + 1.6 g gelatin), B (0.005 g MWCNTs + 1.6 g gelatin), C (0.0075 g MWCNTs + 1.6 g gelatin), and D (0.01 g MWCNTs + 1.6 g gelatin), measured in a UTM by stretching at the rate of 1 mm/min. Reproduced with permission from reference [56]

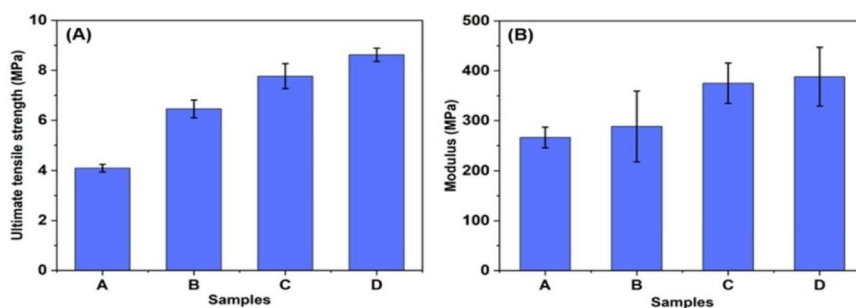
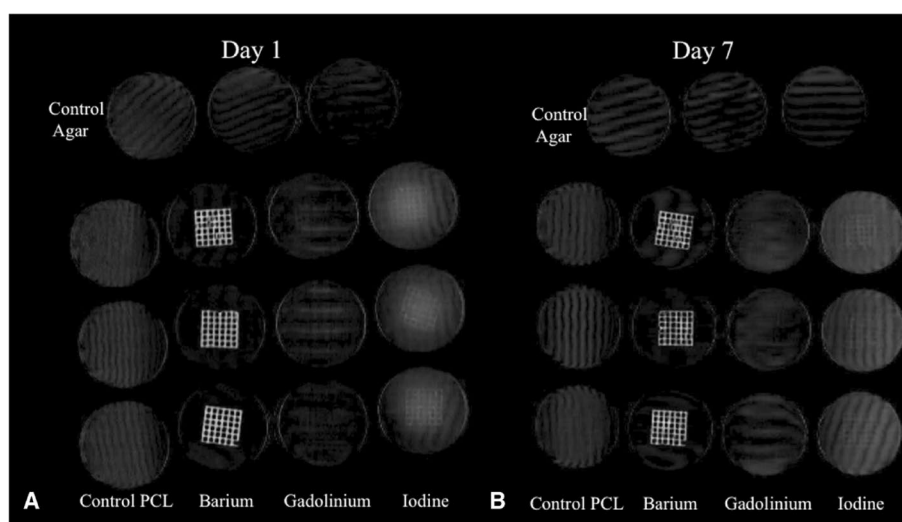


Fig. 5 Coronal volume rendering of the contrast impregnated, and control meshes in the 37 °C agar solution at day 1 (**a**) and day 7 (**b**). Reproduced with permission from reference [58]



These systems must therefore achieve suitable rheological properties for extrusion and adequate shape retention after deposition, frequently through additional stabilization mechanisms such as ionic, covalent, photoinduced, enzymatic, or physical crosslinking [59]. The crosslinking conditions must be carefully controlled, as they may affect printability, mechanical performance, degradation, cytocompatibility, and the activity of incorporated biological components [60].

Furthermore, to increase the biological functionality of printed meshes, several studies have incorporated antibiotics, natural polymers, and biological matrices into these meshes to enhance antimicrobial protection and biological integration. Multifunctionality may be introduced through pre-functionalization, including polymer blending or direct incorporation of active compounds into filaments or inks, or through post-functionalization strategies such as surface coating, plasma treatment, or chemical grafting. The use of antibiotics demonstrated antibacterial activity *in vitro*, though the initial release was too rapid to maintain therapeutic concentrations. In order to overcome this initial burst release, some studies have explored encapsulation strategies using hydrogels, like alginate, to prolong drug release. Meanwhile, the incorporation of biological components, such as gelatin and acellular dermal matrix, has been shown to reduce intra-abdominal adhesion formation and promote cell attachment and tissue regeneration.

Despite the significant progress demonstrated across these studies, several limitations still hinder the clinical application of 3D printing in hernia mesh development. From a mechanical point-of-view, 3D-printed structures (particularly produced by FDM) may fail to replicate the anisotropic behavior and multidirectional load distribution of the abdominal wall. The mentioned initial burst release raises concerns on drug-loaded systems to maintain the therapeutic effect over time. Furthermore, the high temperatures required for thermoplastic extrusion may limit both the use of biopolymers with natural biocompatibility and the functionalization with bioactive compounds, making it more challenging to achieve biologically active interfaces. Low-temperature biomaterial extrusion, in turn, introduces additional challenges related to rheological control and crosslinking.

5 Electrosinning in the production of fibrous mesh structures for hernia treatment

5.1 Fundamentals of electrosinning

Electrosinning is considered a versatile manufacturing technique that uses electrostatic forces to produce fibers with controllable diameters from polymer solutions. The resulting fibers are characterized by their small

diameters, which can range from nanometers to micrometers, their high surface-to-volume ratio, high porosity, ease of surface functionalization, and high flexibility to conform to different shapes and sizes. The technique also allows the use of various polymer solutions, composed of natural and/or synthetic polymers, which may incorporate reinforcing agents and bioactive compounds, thereby enabling the production of structures with advanced properties [61]. In the electrospinning process, the polymer solution is ejected from a needle towards a metallic collector through the application of high voltage. The applied electric field induces charges on the surface of the droplet at the needle tip. As the intensity of the electric field increases, the polymer droplet elongates, acquiring a conical shape commonly referred to as the Taylor cone. When the applied potential overcomes the surface tension of the polymer droplet, a jet of electrically charged fluid is ejected, and the solvent gradually evaporates [61].

5.2 Fibrous mesh structures produced by electrospinning

The biomedical sector particularly benefits from the electrospinning technique for the development of advanced systems used in tissue engineering and drug delivery. One of the main advantages offered by this technique is the flexible selection of materials, as a wide variety of biocompatible and/or biodegradable materials can be easily processed, including synthetic polymers, naturally derived polymers, and nanocomposites. The resulting fibrous architecture promotes cellular activity, as it resembles the ECM of native tissues. The high surface area of the nanofibers facilitates the encapsulation and release of bioactive compounds within therapeutic limits. Scaffold properties, including fiber diameter, porosity, mechanical performance, and degradation resistance, can be tailored by adjusting processing parameters, cross-linking, or incorporating reinforcing materials [62]. Strategies have been developed to enhance the relevance of electrospun materials for tissue engineering applications, leading to the creation of three-dimensional scaffolds with multiscale hierarchical architectures [7].

This technique can also be used to produce structures for hernia repair. Owing to the similarity of these fibrous structures to the ECM, electrospinning shows strong potential for producing the external layer of a multilayered construct designed for hernia treatment. Several studies have already reported the use of electrospinning to produce fibrous structures for hernia repair or for individual layers of such systems. Table 8 summarizes several examples reported in the literature, highlighting material selection, electrospinning equipment parameters, and characterization.

One of the biggest problems found in commercial PP meshes used nowadays is their adhesion to the abdominal wall and the chronic inflammatory response. Regarding this, several studies have been conducted to overcome this limitation by coating the commercial meshes with biocompatible polymers. Xiaolong et al. created a hybrid structure by electrospinning PLA on both sides of a commercial PP mesh [63]. Comparing commercial meshes made of PP or polyester (PE) with the one they developed, the developed mesh showed superior mechanical properties and a significant reduction in intra-abdominal adhesions (Fig. 6), due to PLA's ability to downregulate *TGF- β* expression. Furthermore, the PLA coating promotes a more favorable distribution of collagen I and III, suggesting improved tissue remodeling around the implant. Despite these promising results, a longer in vivo analysis period would be necessary, since after only 12 weeks, it is not possible to predict whether adhesions may occur after PLA degradation. Similarly, Liu et al. electrospun PCL over different knitted PP meshes in order to improve the biological interface of the implant [64]. The results demonstrated that the composite structures exhibited mechanical properties comparable to commercial meshes, and the presence of PCL improved cell adhesion and proliferation. However, the absence of in vivo studies makes it difficult to predict if these results will translate into successful tissue integration and long-term stability within a living organism. Yang et al. and Luan et al. also aim to mitigate adhesions and inflammatory responses by coating commercial PP meshes with the natural polymer silk fibroin [65, 66]. Yang et al. compared their SF-coated mesh with a PCL-coated mesh, demonstrating that the SF layer has a stable degradation over time, unlike the PCL layer [65]. The SF-coated mesh also exhibited active modulation of the immune microenvironment and promotion of angiogenesis, consequently resulting in a lower incidence of long-term adhesions compared to the PCL-coated and PP. Luan et al. obtained similar results, they found that SF promoted the rapid growth of peritoneal mesenchymal cells, significantly

Table 8 Overview of the mesh structures produced by electrospinning, including the materials used, structure and dimensions, processing parameters, and evaluated functional properties

Reference	Parameters	Solutions and concentrations	Evaluated properties
[63]	Flow rate: 0.15 mL/min Voltage: 20 kV Needle-to-collector distance: 10 cm Collector type: rotating	10% (w/w) PLA	Morphological evaluation with SEM Mechanical properties In vitro cytocompatibility and degradation In vivo degradation Intra-abdominal adhesions Histopathology Collagen deposition Pathway analysis
[64]	Flow rate: 25 μ L/min Voltage: 13 kV Needle-to-collector distance: 15 cm Needle: 23 G Collector type: Static Deposition time: 15 min	10% (w/v) PCL	General physical properties Surface morphology characterization Mechanical properties In vitro biocompatibility
[65]	SF: Voltage: 20 kV Needle-to-collector distance: 15 cm Collector type: Static Deposition time: 3 h PCL: Voltage: 18 kV Needle-to-collector distance: 17 cm Collector type: Static Deposition time: 2 h	8.0%, 10.0%, 12.0%, 14.0%, 15.0% and 17.0% (w/v) SF	Fiber morphology Mechanical properties Pore characteristics Cytotoxicity and proliferation In vivo adhesion Histological examination Gene expression
[66]	Flow rate: 4 mL/h Voltage: +15 kV and – 8 kV Needle-to-collector distance: 15 cm Collector type: rotating	10.0% (w/v) SF	Morphological evaluation Structural analysis In vitro cell viability In vivo adhesion Histopathology
[67]	PCL: Flow rate: 1.5 mL/h Voltage: 20 kV Needle-to-collector distance: 20 cm Gelatin: Flow rate: 1.2 mL/h Voltage: 28 kV Needle-to-collector distance: 13 cm PCL/gelatin: Flow rate: 0.5 mL/h Voltage: 21 kV Needle-to-collector distance: 13 cm Needle: 21 gauge Collector type: Static Temperature: RT	12.0% (w/v) PCL 3.0% (w/v) Gelatin 3.68% (v/v) 3-glycidyloxypropyltrimethoxysilane	Morphological evaluation Structural analysis Thermal properties Wettability Hydrolytic degradation Enzymatic degradation Mechanical properties Cytotoxicity/Viability Cell morphology
[68]	Flow rate: 1 mL/h Voltage: 25 kV Needle-to-collector distance: 11 cm Collector type: rotating	1.8%, 2.4% and 5.0% (w/v) PCL 0.6% and 1.2% (w/w) ORC	Structural analysis with FTIR Mechanical properties Morphological evaluation with SEM In vitro stability In vitro cytotoxicity In vivo adhesion Histopathology
[70]	Flow rate: 1 mL/h Voltage: 16–20 kV Needle-to-collector distance: 15 cm Collector type: Static	8.0% (w/w) PVA	Fiber morphology with SEM Mechanical properties Cytotoxicity Cell proliferation In vivo biocompatibility Adhesion formation and tissue ingrowths Histological evaluation

Table 8 (continued)

Reference	Parameters	Solutions and concentrations	Evaluated properties
[71]	Flow rate: 1.5 mL/h Voltage: 22 kV Needle-to-collector distance: 22 cm Needle: 23 G Collector type: Static	12.5% (w/v) PGS 12.5% (w/v) PCL	Morphological evaluation with SEM Degradation behavior Mechanical properties Cytotoxicity evaluation Cell adhesion and proliferation
[72]	Flow rate: 1 mL/h Voltage: 8 kV Needle: 23G	Core: 10.0% (w/w) PU Shell: 5.0% (v/v) PECA	Fiber morphology Physicochemical properties Mechanical properties Tribological analysis Surface properties Biocompatibility Hemocompatibility Cell adhesion
[73]	Flow rate: 10 μ L/min Voltage: 20 kV Needle-to-collector distance: 15 cm Needle: 0.3 mm Collector type: Rotating	85.0% (w/w) PLGA 15.0% (w/w) PLA-b-PEG	Morphological evaluation In vivo anti-adhesion and obstruction Orchiatrophy level Histopathology Cell proliferation Cell apoptosis
[74]	Flow rate: 0.15 mL/min Voltage: 20 kV Needle-to-collector distance: 15 cm Needle: 30 G Collector type: rotating	10.0% (w/w) PLA 1.0% and 2.0% (w/w) Sirolimus	Morphological evaluation Structural analysis Substrate adhesion In vitro cytocompatibility In vivo adhesions Mechanical strength Mesh shrinkage Histopathology and Inflammation
[75]	Flow rate: 1.2 mL/h Voltage: 12 kV Collector type: rotating	9.6% (w/v) PLLA-CL 2.4% (w/v) Gelatine 0.2% (w/v) Ibuprofen	Morphological evaluation Wettability In vivo and ex vivo mechanical properties In vitro degradation Drug loading and release In vitro cytocompatibility In vivo adhesions Histology Inflammatory response
[76]	Flow rate: 2 mL/h Voltage: 10 kV Needle-to-collector distance: 14 cm Needle: 0.60 mm	Core: 10.0% (w/v) PPDO 1.0% (w/v) NAR Shell: 8.0% (w/v) PLLA	Morphological evaluation Structural analysis Wettability Mechanical properties In vitro and in vivo degradation In vitro drug release Histopathology and immunohistochemistry
[77]	Flow rate: 0.5, 1 and 1.5 mL/h Voltage: 2–5 kV (needle) and 10–18 kV (collector)	12.0% (w/w) PCL 1.0% (w/w) Irgasan 0.5% (w/w) Levofloxacin	Rheological characterization Morphological evaluation with SEM and Atomic Force Microscopy (AFM) Wettability measurement In vitro drug release Antibacterial efficacy Chemical distribution of drugs
[78]	Flow rate: 25 μ L/min Voltage: 13 kV Needle-to-collector distance: 10 cm Needle: 23 G Deposition time: 30 min	10% (w/v) PCL 6% (w/v) PEP-1	Morphological evaluation with SEM Structural analysis with FTIR Mechanical properties In vitro drug release and Cytotoxicity Antibacterial activity

Table 8 (continued)

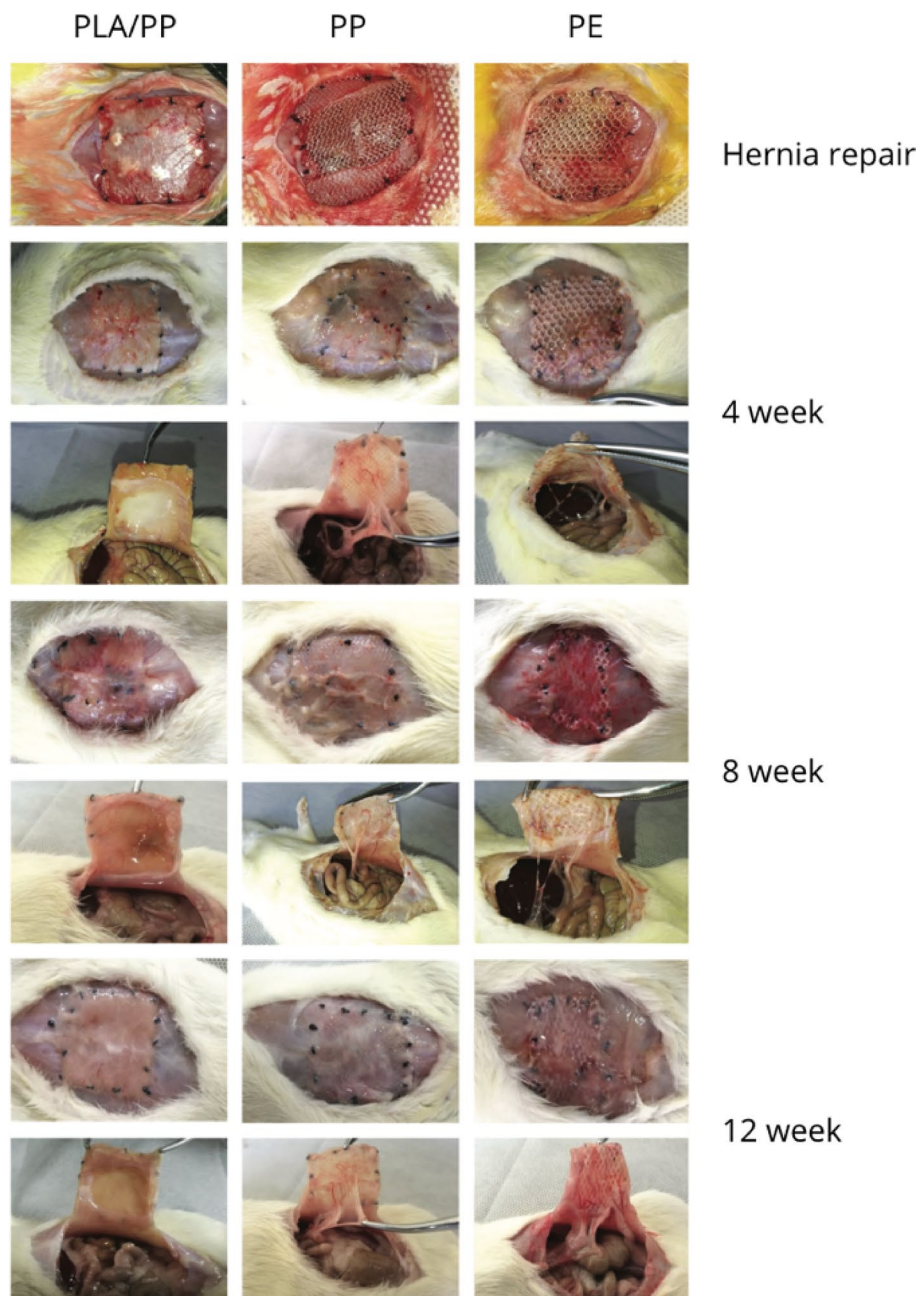
Reference	Parameters	Solutions and concentrations	Evaluated properties
[79]	Flow rate: 1.2 mL/h Voltage: 10 kV Needle-to-collector distance: 15 cm Needle: 1.5 mm Collector type: Static	2.4% (w/v) PCL 3.6% (w/v) SF	Morphological evaluation with SEM Structural analysis Wettability measurement Mechanical properties Amoxicillin release Antibacterial activity In vitro biocompatibility In vivo host response Histological analysis
[80]	Core: Flow rate: 2.5 μ L/min Needle: 0.9 mm Shell : Flow rate: 5 μ L/min Needle: 0.57 mm Voltage: 22 kV Needle-to-collector distance: 20 cm Humidity: 15–20% Temperature: 20–22 °C Collector type: Rotating Deposition time: 4–6 h	Core: 21.0% (w/v) PA6 0.3% (w/v) PHMB Shell: 2.40% (w/v) CS 0.60% (w/v) PEO 0.45% (w/v) 5-chloro-8-hydroxyquinoline	Fiber morphology Transmission electron microscopy (TEM) Structural analysis Wettability measurement Mechanical properties In vitro drug release Antimicrobial activity
[82]	Flow rate: 0.5–1.5 mL/h Voltage: 13–16 kV Needle-to-collector distance: 20 cm Needle: 21G Collector type: rotating	7.0%, 9.0%, 10.0% and 12.0% (w/w) PCL	Morphological evaluation Wettability Surface chemistry Cell viability Anticoagulant activity Hemocompatibility
[83]	PCL/PTBAEMA Flow rate: Core: 0.2 mL/h Shell: 0.4 mL/h PCL/PAMPS Flow rate: Core: 0.2 mL/h Shell: 0.1 mL/h Voltage: 22 kV Needle-to-collector distance: 20 cm Needle: 21 G Collector type: rotating	20.0% (w/w) PAMPS 18.0% (w/w) PTBAEMA 12.0% (w/w) PCL	Morphological evaluation Structural analysis Wettability Thermal properties Aqueous stability In vitro cell viability In vitro anticoagulant activity In vitro antibacterial activity
[84]	Flow rate: 1.5 mL/h Voltage: 15 kV Needle-to-collector distance: 12 cm Needle: 20G Collector type: static	4.5%, 9.0%, and 13.5% (w/v) SF 4.5%, 9.0%, and 13.5% (w/v) PHBV	Morphological evaluation with SEM Structural analysis with FTIR Thermal properties Mechanical properties In vitro cytocompatibility Gene expression In vivo tissue regeneration and inflammatory response
[85]	Flow rate: 0.6 mL/h Voltage: 18 kV Needle-to-collector distance: 10 cm Needle: 20G Collector type: Static	2.0% (w/v) SF 8.0% and 10.0% (w/v) PLA	Morphological evaluation Structural analysis Wettability Mechanical properties In vitro swelling and degradation In vivo degradation In vitro cell behavior In vivo adhesions Histopathology Tissue repair and angiogenesis

Table 8 (continued)

Reference	Parameters	Solutions and concentrations	Evaluated properties
[86]	PCL 15%: Flow rate: 3 mL/h Needle-to-collector distance: 12 cm PCL 20%: Flow rate: 5 mL/h Needle-to-collector distance: 14 cm Chitosan: Flow rate: 1 mL/h Needle-to-collector distance: 14 cm Voltage: 20 kV Needle: 20 G Collector type: rotating Humidity: 25–30% Temperature: 25–27 °C	15.0% and 20.0% (w/w) PCL 2.5% (w/w) CS 9.1% (w/w) PEO	Morphological evaluation Structural analysis Mechanical properties Wettability Swelling and degradation In vitro cellular behavior In vivo adhesions Histopathology and fibrogenesis
[87]	Flow rate: 2 mL/h Voltage: 15.5 kV Needle-to-collector distance: 20 cm Deposition time: 30 min	20.0% (w/v) PCL	Adhesion Hernia recurrence Histopathology Immunohistochemistry
[88]	Flow rate: 1.2 mL/h Voltage: 23 kV Needle-to-collector distance: 20 cm Collector type: Static	2.50%, 2.25%, 2.00% and 1.75% (w/v) CS 0.25%, 0.50% and 0.75% (w/v) ES	Morphological evaluation with SEM Wettability and swelling Mechanical properties In vitro degradation Antibacterial activity In vitro cytocompatibility In vivo tissue regeneration Histology
[89]	Flow rate: 0.4 mL/h Voltage: 15 kV Needle-to-collector distance: 12 cm Needle: 0.8 mm Collector type: Static Humidity: >60% Temperature: 25–28 °C	30.0% (w/w) SF	Morphological evaluation with SEM Structural analysis Component analysis Biocompatibility Bioactivity In vitro degradation In vivo adhesion Histological analysis Mechanical properties Molecular mechanism
[90]	Flow rate: 1.2 mL/h Voltage: 15 kV Needle-to-collector distance: 12 cm Needle: 0.8 mm Humidity: 60% Temperature: 25–28 °C	3.6% (w/v) PCL 2.4% (w/v) SF 0.6% (w/v) HAM	Morphological evaluation with SEM Structural analysis Wettability In vitro degradation In vitro and ex vivo mechanical properties Cytocompatibility Bioactivity In vivo adhesions Histological analysis

attenuated the local inflammatory response, and actively stimulated neovascularization [66]. Giuntoli et al. used a blend of PCL and gelatin type A to obtain a mesh that better mimics the native ECM [67]. The results indicated that the developed meshes maintained mechanical properties while promoting fibroblast attachment. However, the authors observed a reduction in cell metabolism and a mild cytostatic effect during the initial 72 h of culture. Although this effect seemed to diminish over time, it suggests that the initial cellular response to the material might be temporarily affected, which could influence early tissue integration. Sezer et al. explored a similar strategy as the other authors by modifying the mesh surface with electrospun layers of PCL and ORC [68]. The authors concluded that the coated meshes maintained adequate mechanical properties while improving surface characteristics that favor cell attachment and proliferation. In addition, the presence of ORC was suggested to

Fig. 6 Hernia repair with three kinds of mesh and observation of the adhesions 4, 8 and 12 weeks after surgery in the three groups. Reproduced with permission from reference [63]



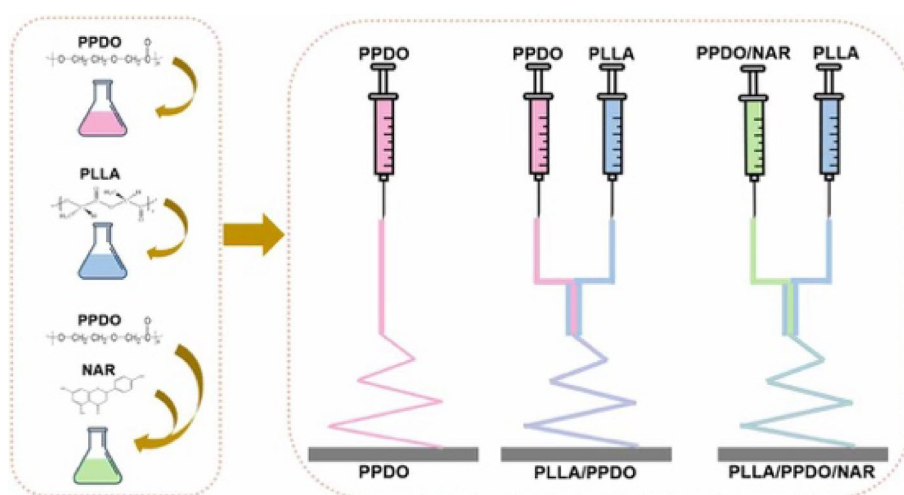
contribute to the reduction of inflammatory responses and adhesion formation due to its bioresorbable nature and its ability to act as a temporary protective barrier during the initial healing phase, besides the known positive effect on mechanical properties in PCL matrices provided by nanoscale cellulose [69].

Besides coatings, researchers have developed independent electrospun scaffolds designed specifically as physical barriers. Molnár et al. investigated the use of cross-linked PVA to engineer a non-adhesive, absorbable scaffold [70]. The results showed that while its mechanical forces were lower than those of standard PP meshes, it remained suitable for small hernia repair. The in vivo analysis revealed a structural integrity of the scaffold after approximately 6 months, with no evidence of several adhesions, chronic inflammation, or bacterial film formation. Kaya et al. produced a bilayer mesh with a non-degradable and non-adhesive PU layer facing the viscera and an absorbable poly (glycerol sebacate)/poly(caprolactone) (PGS/PCL) layer facing the wall to support healing

[71]. Although the PGS component degraded rapidly within the first 15 days, the mesh remained highly elastic (10 times the control) and effectively prevented adhesions and infection while supporting endothelial cell growth. Chalony et al. developed a core-shell mat combining poly (ethyl-2) cyanoacrylate (PECA) and PU [72]. The final mat exhibited improved mechanical resistance and elongation when compared to pure PU, however the researchers found that the mat gradually lost structural strength over time. Biologically, it showed attenuation of fibroblast and platelet adhesion. Liu et al. inserted an electrospun fibrous membrane made of poly(lactic-co-glycolic acid)/poly(lactide)-b-poly(ethylene glycol) (PLGA/PLA-b-PEG) between a PP mesh and the spermatic cord to act as a temporary physical barrier [73]. While the PP mesh by itself induced a severe inflammatory response, resulting in complete vas deferens obstruction, dense tissue adhesions, and an over 30% reduction in testicular weight, the introduction of the electrospun barrier successfully mitigated early-stage inflammation and preserved normal testicular size and function. The membrane completely degraded within 180 days but successfully prevented early adhesions, allowing newly formed host tissue to encapsulate the polypropylene mesh.

Beyond physical barriers, electrospinning has been extensively investigated as a platform for localized pharmacological delivery to actively modulate the host response. Zhang et al. developed a composite mesh by electrospinning a sirolimus-loaded PLA membrane directly onto a commercial PP mesh, utilizing the drug's immunosuppressive and anti-proliferative properties to actively prevent adhesion formation instead of just providing a physical barrier [74]. Despite achieving smaller adhesion areas compared to control groups, a high dose of sirolimus compromised systemic wound healing, leading to hernia recurrence. This finding suggests that future designs must carefully balance between preventing visceral adhesions and promoting fascial repair, because the mechanisms that stop fibrosis also significantly inhibit the vascularization and collagen deposition needed for long-term mechanical stability. Liu et al. engineered a bilayer electrospun mesh made of poly(L-lactide-co-ε-caprolactone) (PLLA-CL) and gelatin loaded with ibuprofen [75]. The bilayer structure presented mechanical properties suitable for in vivo repair, while the sustained release of ibuprofen successfully shifted macrophage polarization from a pro-inflammatory M1 toward a pro-remodeling M2 phenotype, resulting in a mesh that remained largely adhesion-free while facilitating robust infiltration of mature blood vessels and organized collagen I deposition. However, the early burst release of approximately 70% of the drug within 4 days suggests that the immunomodulatory effect is concentrated in the immediate postoperative period. A study conducted by Xv et al. used a coaxial electrospinning technique to construct a core-shell nanofibrous delivery system [76]. To achieve a more prolonged immunomodulatory effect, the researchers encapsulated naringin within a poly(p-dioxanone) (PPDO) core, which was protected by a PLLA sheath (Fig. 7). The use of this technique drastically reduced the initial burst release, enabling a sustained delivery extended over 19 days. In vivo studies demonstrated that the use of naringin significantly reduced pro-inflammatory interleukin-6 expression, effectively protecting the tissue from chronic inflammation, and also improved the amount of collagen deposited.

Fig. 7 Schematic illustration of the core-shell nanofiber structure. Reproduced with permission from reference [76]

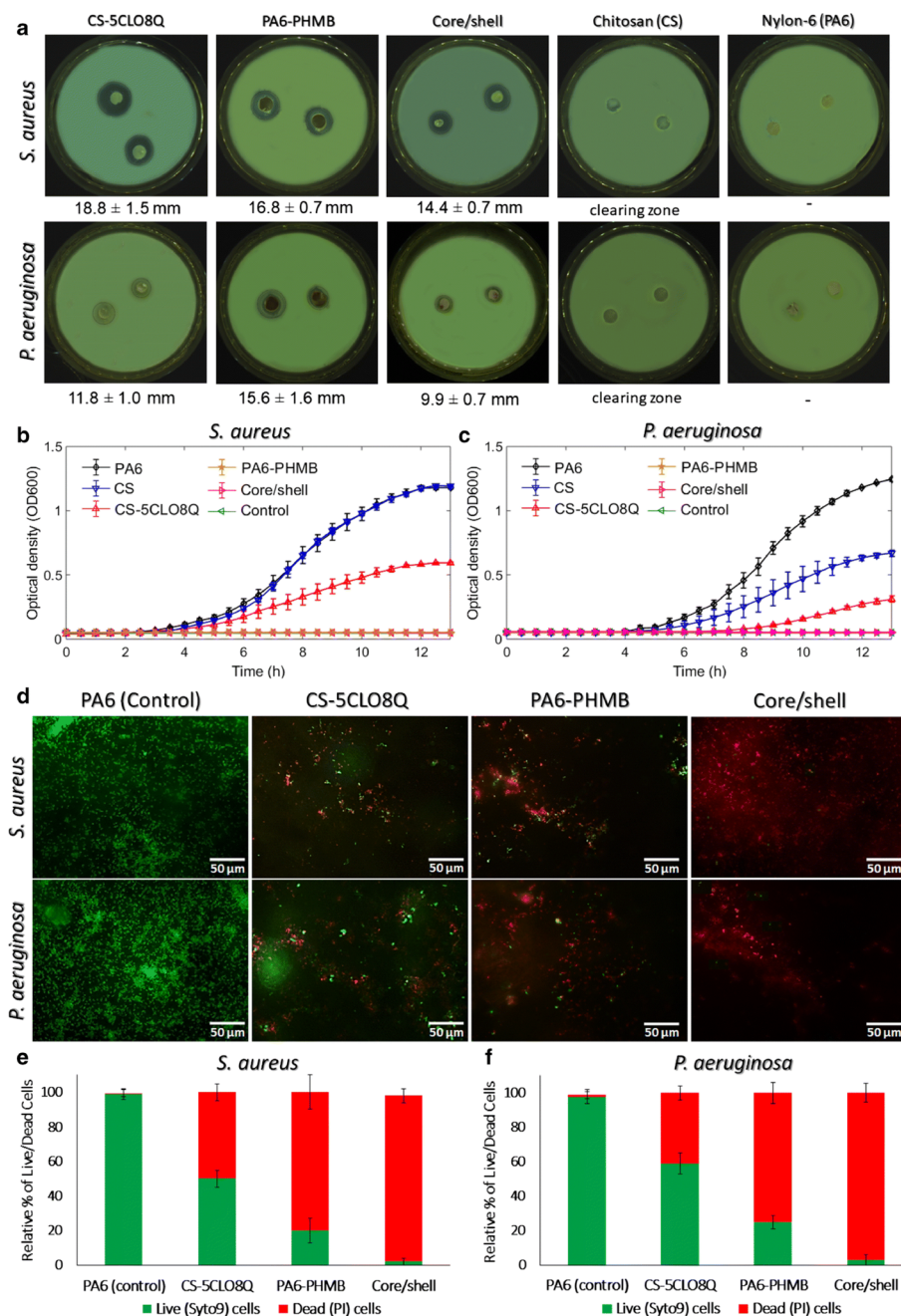


Another application of electrospun lies in infection control, where researchers have developed scaffolds for localized antimicrobial delivery to prevent surgical site infections. Hall Barrientos et al. produced PCL electrospun scaffolds loaded with either the antibacterial agent igrasan or the antibiotic levofloxacin [77]. The results revealed that the antimicrobial agent was successfully incorporated into the internal polymeric matrix, enabling a highly sustained release over 145 h that provided continuous bacterial inhibition. In contrast to the controlled release observed previously, the antibiotic scaffold exhibited surface crystallization, leading to a rapid burst release within the first 15 min. Similarly, Liu et al. used electrospun PCL loaded with an antimicrobial peptide (PEP-1) to functionalize a commercial PP mesh [78]. The produced mesh exhibited an initial burst release of approximately 25% within the first 90 min, followed by a slow release reaching 48% over 28 days. While the mesh successfully inhibited the Gram-negative *E. coli*, it failed to inhibit *S. aureus*, one of the most prevalent pathogens responsible for mesh infections. In a subsequent study, Liu et al. engineered a composite mesh by electrospinning a blend of PCL and SF, and subsequently chemically grafted amoxicillin-loaded multiwalled carbon nanotubes (MWCNTs) directly onto the nanofiber surfaces [79]. The authors concluded that functionalization improved the elasticity of the PCL/SF structure, reduced the initial burst release of amoxicillin by enabling sustained release, and decreased inflammatory and foreign-body reactions in vivo. However, as discussed in Sect. 4.2 regarding the study by Yadav et al., the use of MWCNTs may imply relevant concerns regarding cytotoxicity and biocompatibility.

To overcome the need for potentially toxic nanomaterials or complex drug-encapsulation techniques, some research groups have developed strategies that shift toward using polymers with intrinsic bioactive properties. Keirouz et al. combined Polyhexanide (PHMB), Nylon-6, 5-chloro-8-quinolinol, and chitosan in a coaxial electrospinning setup to create core-shell nanofibers [80]. The resulting scaffold exhibited a sustained release with sufficient bactericidal activity against both Gram-negative and Gram-positive bacteria (Fig. 8). The scaffold also achieved mechanical properties significantly superior to those of its individual constituent polymers, due to the energy dissipation caused by the phase-separated “beads-in-fiber” morphology, which has been shown to enhance the toughness of core-shell nanofibrous materials and increase tensile strength and Young’s modulus [81]. Dufay et al. used cold plasma treatment to chemically graft poly (2-acrylamido-2-methylpropane sulfonic acid) (PAMPS) onto the surface of PCL nanofibers [82]. This surface modification was made to mimic the anticoagulant activity of heparin by inhibiting fibrin deposition, which fundamentally drives the formation of visceral adhesions. The authors concluded that the functionalization delayed the clotting cascade, thereby preventing the early fibrin networks that serve as scaffolds for adhesion formation. In 2025, Dufay et al. developed a bilayer mesh via coaxial electrospinning, with PCL and poly((2-tert-butylamino) ethyl methacrylate) (PTBAEMA) on one side and PCL and PAMPS (reported in the previous article) on the other, providing antimicrobial and anticoagulant properties, respectively [83]. The results showed a massive and sustained bactericidal effect, with the complete eradication of colonies of both Gram-positive and Gram-negative bacteria in 24 h.

Advanced designs have explored blending synthetic polymers with extracted natural proteins to promote tissue regeneration actively by mimicking the natural ECM. For instance, Gong et al. blended SF with poly(3-hydroxybutyrate-co-3-hydroxyvalerate) (PHBV), demonstrating that increasing the proportion of the natural silk protein improved the elongation at break and porosity [84]. Furthermore, the inclusion of SF facilitated fibroblast adhesion, increased collagen deposition, and simultaneously mitigated strong inflammatory responses. Wang et al. electrospun PLA and SF, reaching values of tensile strength four times higher than those found in the commercial PP meshes [85]. This scaffold also increased fibroblast proliferation and migration, as well as neovascularization, thereby preventing adhesions and causing only mild inflammatory responses. However, the high tensile strength obtained raises concerns, as it may not provide sufficient flexibility for use in the abdominal wall. Asvar et al. engineered an electrospun bilayer scaffold composed of PCL on the side of the viscera to act as a physical barrier against adhesions, and of CS on the side of the wall to actively promote tissue integration [86]. The addition of CS increased the attachment, spreading, and metabolic viability of fibroblasts and endothelial cells in vitro, while

Fig. 8 Antimicrobial performance of electrospun mats against *S. aureus* and *P. aeruginosa*. **(a)** Representative disc-diffusion assays and corresponding zones of inhibition (mean $\pm$ SD) for CS–5CLO8Q, PA6–PHMB, core–shell, chitosan (CS), and Nylon-6 (PA6) mats. Bacterial growth curves of **(b)** *S. aureus* and **(c)** *P. aeruginosa*, monitored by optical density at 600 nm (OD₆₀₀). **(d)** Representative fluorescence microscopy images obtained using the BacLight live/dead assay, with live and dead bacteria stained green and red, respectively. Relative percentages of live and dead cells after exposure to the different electrospun mats for **(e)** *S. aureus* and **(f)** *P. aeruginosa*. Reproduced with permission from reference [80]



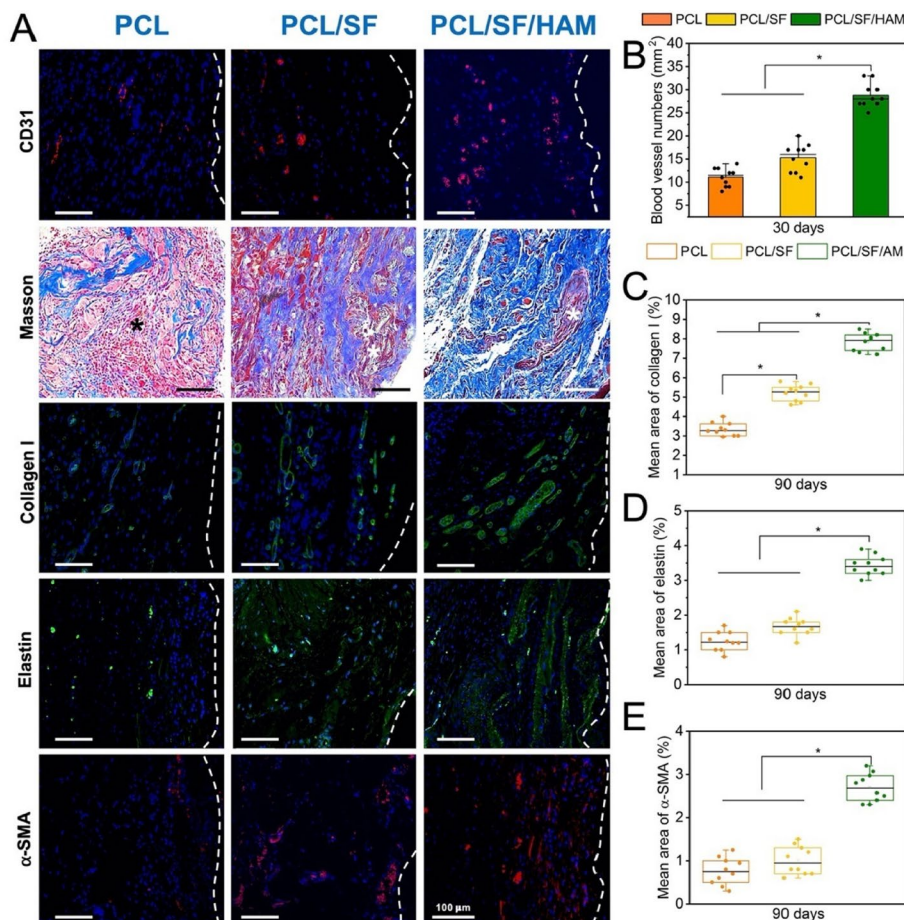
the final scaffold successfully prevented macroscopic adhesions *in vivo* and achieved strain values comparable to those found in the native abdominal wall.

Beyond hybrid scaffolds, some approaches use decellularized membranes to preserve native ECM complexity. However, Scuglia et al. demonstrated their mechanical weakness by comparing an electrospun PCL membrane and a biological decellularized human chorion membrane (dHCM) with Gore-Tex® (PTFE mesh) in a growing porcine diaphragmatic hernia model [87]. The results showed that, despite superior tissue integration and neo-vascularization of PCL and dHCM, their rapid *in vivo* degradation under physiological tension led to 50% and 100% recurrence rates for PCL and dHCM, respectively. The Gore-Tex® had a 0% recurrence rate, although it presented dense fibrotic encapsulation.

To overcome the rapid degradation of biological matrices and physically reinforce intact biological membranes, some efforts have been made. Cao et al. electrospun a solution of CS and ES directly onto the surface of porcine small intestinal submucosa (SIS) [88]. The results showed that the addition of CS and ES reinforced the matrix, actively maintaining its micro-stiffness under dynamic loading. Furthermore, this biocomposite facilitated cellular infiltration, stimulated angiogenesis, and exhibited antibacterial activity, inhibiting the growth of *E. coli* and *S. aureus*. Liu et al. used an electrospun SF membrane to reinforce a decellularized human amniotic membrane (HAM) [89]. The developed biocomposite presented higher concentrations of angiogenic factors compared to commercial SIS. In vivo, the composite showed only weak localized adhesions, and the SF reinforcement successfully matched the mechanical strength of the commercial mesh by day 28. Liu et al. also investigated the use of SF and HAM with PCL. In this study, they blended all the components and electrospun them to create a PCL/SF/HAM membrane [90]. The resulting membrane demonstrated mechanical elongation that approximated native abdominal wall tissue, and the addition of HAM provided a better microenvironment for cell proliferation. In vivo results revealed fewer adhesions along with histologically weaker inflammatory responses and foreign body reactions. During remodeling, the bioactive structure induced more pronounced neovascularization and remarkable incorporation of collagen and elastin fibers (Fig. 9).

The studies reviewed demonstrate that electrospinning offers a versatile platform for addressing the major complications of hernia repair, such as postoperative adhesions, chronic inflammation, infection, and poor tissue integration. Several strategies emerged from this advanced fabrication approach. Coating commercial PP mesh with synthetic biodegradable polymers has shown promising results. Xiaolong et al. demonstrated that PLA coatings reduce adhesion formation and improve collagen organization, while Liu et al. reported similar

Fig. 9 Histological and immunofluorescence evaluation of PCL, PCL/SF, and PCL/SF/HAM meshes after implantation. (A) CD31 staining at the centre of the explants after 30 days, and Masson's trichrome, collagen I, elastin, and α -SMA staining after 90 days. Quantitative analysis of (B) blood vessel density, (C) collagen I, (D) elastin, and (E) α -SMA expression. Scale bars represent 100 μ m. The asterisk in panel A indicates undegraded scaffold material. $P < 0.05$. Reproduced with permission from reference [90]



improvements using PCL, which also promoted cell adhesion and proliferation [63, 64]. Furthermore, as shown by Yang et al. and Luan et al., the use of natural polymers, such as silk fibroin, in the coating can significantly enhance this reduction by actively modulating the immune microenvironment, promoting angiogenesis, and improving both biological integration and mechanical properties of the patch [65, 66]. Beyond surface modification strategies, fully electrospun patches have been proposed as alternatives to permanent PP meshes. This approach demonstrates promising biological performance, however, mechanical limitations remain a challenge for fully electrospun structures intended to withstand abdominal wall forces. Consequently, electrospinning is predominantly explored as a complementary strategy, often combined with mechanically robust supports rather than used as a complete replacement for conventional mesh structures.

Beyond their use as coatings or physical barriers, electrospun fibers can be multifunctionalized through pre-functionalization strategies, including polymer blending, incorporation of drugs, nanoparticles, natural polymers, or other bioactive compounds into the spinning solution, as well as through post-functionalization approaches such as plasma treatment, chemical grafting, surface coating, and impregnation. These strategies may allow the use of electrospun membranes as a localized, sustained pharmacological delivery to combat infection and chronic inflammation without the risks of systemic toxicity. The use of broad-spectrum antibiotics, such as levofloxacin and amoxicillin, immunosuppressants like sirolimus, naturally derived flavonoids like naringin, and anti-inflammatory drugs such as ibuprofen have demonstrated the potential to inhibit surgical pathogens, limiting macrophage-driven inflammation, and preventing excessive or disorganized collagen deposition. Furthermore, instead of depending only on traditional antibiotic loading, which can cause burst-release toxicity and promote antibiotic-resistant bacteria, researchers have also reported antibacterial activity using polymers with natural bioactive properties. Nevertheless, this does not necessarily translate into the prevention of surgical site infection *in vivo*. Host response should be evaluated through complementary outcomes, including inflammatory and foreign-body responses, collagen organization, vascularization, adhesion formation, and tissue remodelling, while considering the limitations and duration of the animal model used. Materials with inherent positive charges, such as chitosan or specific amine-functionalized polymers like PTBAEMA, actively break down bacterial membranes, providing strong, sustained antimicrobial protection without depending on conventional antibiotics. Conventional electrospun drug-delivery systems may also present an initial burst release, potentially reducing the duration of the therapeutic effect. To improve these therapeutic delivery systems, researchers have increasingly adopted advanced fabrication techniques such as coaxial electrospinning. By creating specialized core-sheath architectures, this multi-scale approach protects sensitive bioactive molecules, prevents premature drug release, and enables complex biphasic therapeutic systems, such as the dual-delivery of PHMB and 5-chloro-8-quinolinol from Nylon-6/chitosan fibers developed by Keirouz et al. to provide sustained protection against severe surgical site infections.

While functionalizing synthetic materials represents a major focus in the field, another critical area of research addresses the limitations of purely biological meshes. Synthetic polymers were predominant among the reviewed studies, with PCL, PLA, PLLA-CL, PLGA, PEO, and PVA being the most frequently reported materials, due to their good processability, reproducibility, and greater control over mechanical and degradation properties. Although fibers composed exclusively of natural polymers can be produced, their fabrication is often limited by poor spinnability and insufficient mechanical strength. Therefore, natural polymers such as silk fibroin, gelatin, chitosan, and elastin are commonly blended with synthetic polymers. Although naturally derived scaffolds, such as SIS and HAM, present superior immunomodulatory and angiogenic properties, they are highly susceptible to rapid enzymatic degradation, leading to premature mechanical failure. This limitation was highlighted by Scuglia et al., who reported increased hernia recurrence rates when purely biological membranes were tested in demanding large-animal models [87]. To address this, some groups have explored reinforcement strategies, such as Cao et al., reinforcing SIS with electrospun chitosan and elastin, or Liu et al. using silk fibroin to stabilize HAM, achieving improved mechanical stability while preserving bioactivity [88, 90]. However, these materials still present additional clinical challenges, including potential risks of disease transmission, donor-to-donor variability, and ethical considerations.

Despite the considerable advantages of electrospinning when it comes to biological interactions and mimicking the ECM, there are still several limitations that contain its application as a solution for hernia mesh development. The main challenge is related to the insufficient mechanical properties of electrospun fibers, commonly unable to withstand the abdominal wall load. Their low load-bearing capacity requires combination with a mechanically robust mesh. Electrospinning offers a versatile platform for bioactive agent incorporation, although conventional electrospun systems may present an initial burst release that limits their long-term effectiveness. Advanced configurations, such as coaxial electrospinning, can provide greater control over release kinetics, but their processing complexity may hinder reproducibility and scale-up. Additionally, the lack of design personalization can limit their ability to reproduce the complex architecture of the abdominal wall. Moreover, the variability in the processing parameters can affect the reproducibility and scalability of these solutions when it comes to large-scale production. These limitations highlight the need for hybrid strategies that combine the biological advantages of electrospinning with mechanically robust fabrication techniques.

6 Combination of 3D printing and electrospinning in tissue engineering

The combination of 3D printing and electrospinning represents a promising strategy to overcome the limitations associated with each technique individually. While 3D printing enables precise control over macroscopic architecture, internal geometry, and pore distribution, electrospinning provides high surface area, nanofibrous morphology, and structural similarity to the ECM. Integrating both techniques allows the fabrication of scaffolds that combine mechanical stability with enhanced cell adhesion, nutrient diffusion, and cellular integration (Fig. 10).

The incorporation of absorbable synthetic polymers into both 3D printing and electrospinning has been increasingly explored, as these polymers degrade after fulfilling their reconstructive function, thereby avoiding foreign-body reactions [5]. Absorbable synthetic polymers processed by 3D printing are considered an innovative approach. Their degradability, combined with the precision and customization offered by 3D printing, enables the production of tailored structures [91]. Their high porosity and high surface-to-volume ratio promote cell proliferation, making them promising candidates for biomedical applications. The advantages of electrospinning,

Fig. 10 Schematic representation of the complementary roles of 3D printing and electrospinning in the development of hybrid next-generation hernia meshes. The 3D-printed layer provides load-bearing support and architectural control, while the electrospun layer acts as a tissue-interfacing and drug-delivery component

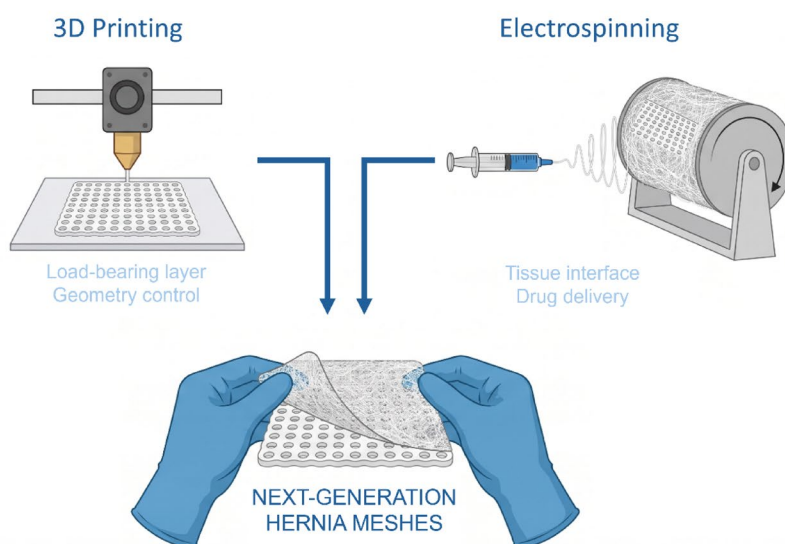
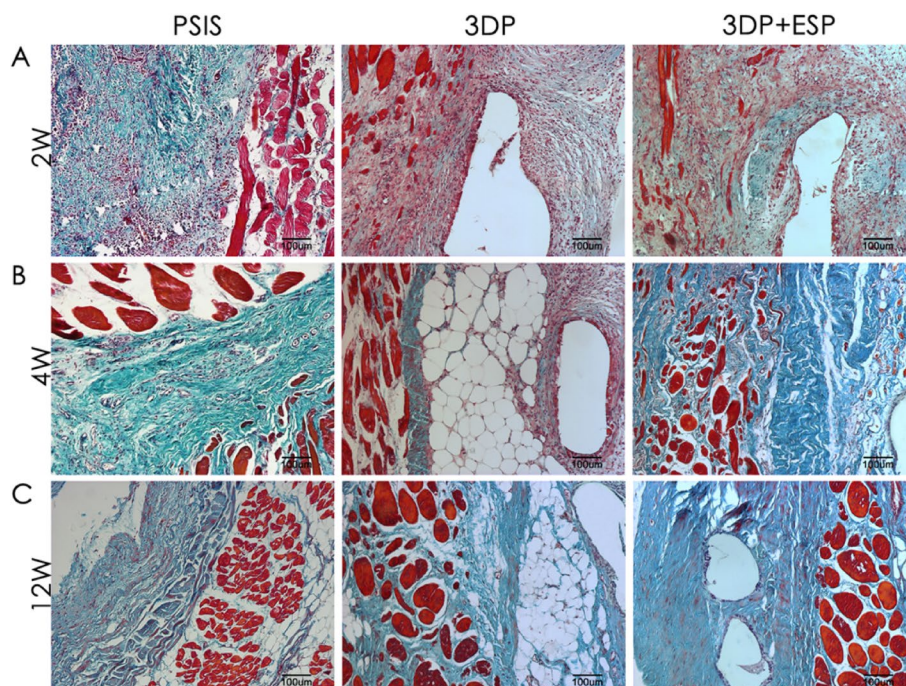


Fig. 11 Masson's trichrome staining of PSIS, 3DP, and biological composite scaffolds during repair of an abdominal wall defect 2 weeks (2 W), 1 month (1 M), and 3 months (3 M) after implantation. Reproduced with permission from reference [92]



combined with the degradability and biocompatibility of biodegradable polymers, make this approach particularly attractive [10].

According to Smith and collaborators, two main strategies have been reported for combining electrospinning with 3D printing [7]. The first involves the layer-by-layer deposition of electrospun nanofibers and 3D-printed elements, resulting in multilayer architectures. The second involves 3D printing inks that contain electrospun nanofibers, referred to as composite inks. More recent studies have explored the use of uniaxially aligned electrospun fibers to functionalize 3D-printed surfaces, thereby improving cell adhesion, proliferation, and differentiation [7]. Table 9 presents reported cases combining 3D printing and electrospinning for the fabrication of fibrous structures.

Yang et al. produced a smart scaffold by alternating electrospun layers with 3D-printed layers [92]. The 3D-printed layers were made with PLLA-CL to serve as a macrostructural support, and the electrospun layers were made with collagen type I to act as a biomimetic ECM. The results showed a highly porous and interconnected architecture capable of efficient cell entrapment. In vivo assays revealed early neovascularization, minimal intra-abdominal adhesions, organized collagen alignment, and improved biomechanical incorporation by twelve weeks compared to both pure printed scaffolds and commercial biological mesh (Fig. 10).

Hu et al. designed a composite patch using 3D-printed PCL for structural support, combined with an electrospun anti-adhesive layer of PVA and a soybean peptide (SP) [29]. The developed patch showed superior elongation at break compared with commercial PP meshes, with a maximum tensile strength that exceeded the empirically derived reference values reported for abdominal wall repair under the tested conditions. The incorporation of SP increased endothelial proliferation and scaffold coverage by day 5. In vivo, the patch demonstrated lower adhesions than the PCL structure, promoted neovascularization, and supported myofibroblast tissue formation. Devenci et al., aiming to combat surgical site infections and wound exudates, engineered a biopatch using a combination of 3D printing and coaxial electrospinning [9]. To provide temporary mechanical support and localized antibacterial therapy, they used a 3D-printed PCL and gelatin loaded with the antibiotic CIP. For the electrospun layer, they used PCL, gelatin, and κ -carrageenan (κ -C) to mimic the ECM. The authors concluded that the PCL matrix successfully prevented the burst release of CIP, achieving controlled release for over 28 days with bactericidal

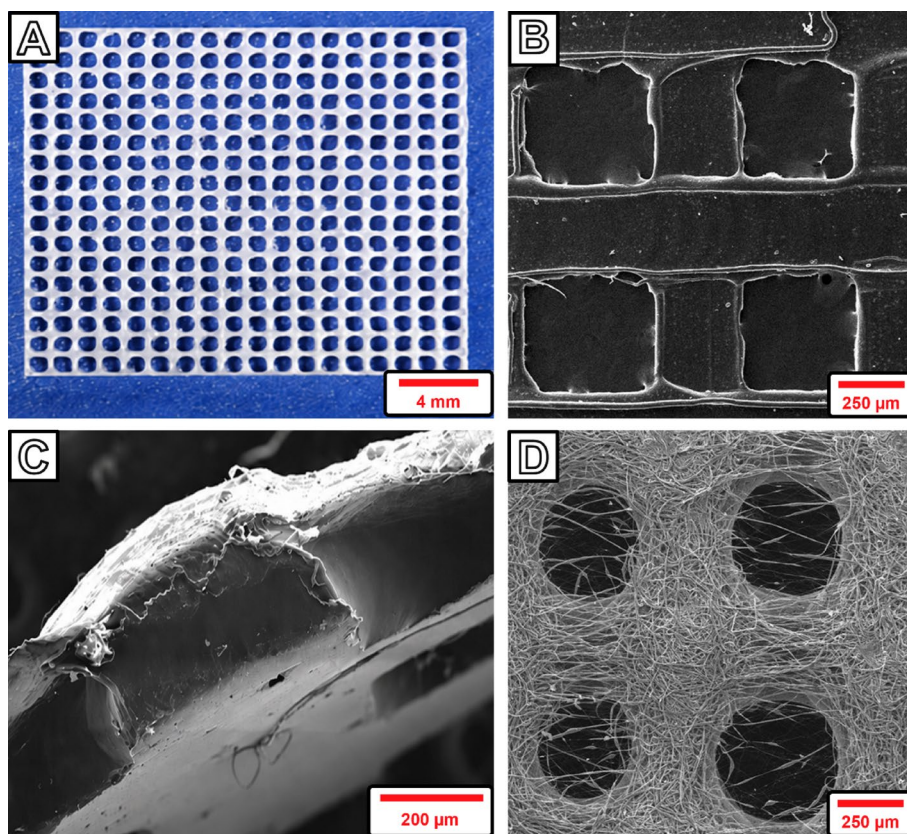
Table 9 Overview of the hybrid mesh structures produced by 3D printing and electrospinning, including the materials used, structure and dimensions, processing parameters, and evaluated functional properties

Reference	Materials	Electrospinning parameters	3D printing parameters	Evaluated functional properties
[92]	PLLA-CL Collagen I	Flow rate: 0.39 mL/min Voltage: 15 kV Needle-to-collector distance: 15 cm Needle: 0.9 mm	Nozzle Temperature: 190 °C Deposition speed: 300 mm/min Layer height: 225 µm	Morphological evaluation In vivo tests In vitro cell proliferation tests Macroscopic observations Histopathology Biomechanical properties
[29]	PCL PVA SP	Flow rate: 0.8 mL/h Voltage: + 15 and – 5 kV Needle-to-collector distance: 10 cm Needle: 22G Collector type: Static Humidity: 30% Temperature: 25 °C	Extrusion pressure: 2 kg/cm ² Needle: 22G	Morphological evaluation Structural analysis Wettability Mechanical properties In vitro cell viability, adhesion, and proliferation In vivo adhesions Tissue regeneration
[9]	PCL Gelatine κ-C CIP	Core: Flow rate: 0.4 mL/h Voltage: 22.5 kV Needle-to-collector distance: 12 cm Needle: 2.13 mm Deposition time: 2 h Shell: Flow rate: 2.5 mL/h Voltage: 22.5 kV Needle-to-collector distance: 12 cm Needle: 3 mm Deposition time: 2 h	Nozzle Temperature: 23 °C Nozzle: 28 G Pressure: 10 psi Printing speed: 80 mm/s Layer height: 0.105 mm Number of layers: 7	Morphological evaluation Structural and rheological analysis Thermal properties Mechanical properties In vitro degradation, swelling, and drug release Antibacterial activity Cell viability Wound healing Histopathology Biochemical analysis
[93]	PCL GelMA SA Van	Flow rate: 0.45 mL/h Voltage: +8 and – 1.5 kV Needle-to-collector distance: 15 cm Needle: 22G Humidity: 30% Temperature: 25 °C	Not reported	Morphological evaluation Mechanical properties Wettability In vitro swelling, degradation, and drug release Antibacterial properties In vivo adhesions and tissue regeneration

activity against Gram-negative and Gram-positive bacteria. Furthermore, κ-carrageenan presented high swelling capabilities, allowing the patch to absorb wound exudate. In vivo results indicated neither hernia recurrence nor severe intra-abdominal adhesions. Moreover, the biopatch elicited significantly less inflammatory cell infiltration and actively protected the repairing tissue from oxidative stress compared to commercial PP meshes. Hu et al. investigated the combination of three advanced manufacturing techniques, such as electrospinning, coaxial 3D printing, and electrospray, to develop a bilayered composite patch [93]. To provide mechanical support, prevent infection, and promote tissue integration, a coaxial 3D-printed layer was created with PCL, gelatin methacryloyl (GelMA), sodium alginate (SA), and the antibiotic vancomycin (Van). Conversely, to create a physical barrier on the visceral side, they electrospun PCL nanofibers and subsequently applied electrostatic spraying to deposit silicon dioxide (SiO₂) microspheres onto the surface. In vitro assays demonstrated that the modified surface properties of the PCL–SiO₂ layer reduced unwanted cellular adhesion, while the highly hydrophilic GelMA/SA exhibited significant swelling capacity to manage wound exudate, promoted endothelial cell proliferation, and provided sustained vancomycin release to provide antibacterial activity against surgical pathogens such as *S. aureus* and *E. coli*. In vivo studies revealed no intra-abdominal adhesions, while histological analysis indicated that the biopatch actively suppressed pro-inflammatory markers and upregulated pro-healing factors.

In conclusion, the findings of the reviewed studies illustrate the complementary roles of the two fabrication techniques. Their combination brings together the strengths of each to produce fibrous scaffolds with enhanced

Fig. 12 Macroscopic and micro-morphology of hybrid 3D-printed/electrospun structures. (A) Macroscopic view of PCL 3D-printed porous mesh. (B) SEM image of the PLLA-CL 3D-printed pore architecture. (C) Cross-sectional SEM image showing the electrospun layer deposited onto the printed support of the PCL-PVA/SP composite mesh. (D) Surface SEM image of the PLLA-CL/Collagen hybrid structure, showing the nanofibrous network covering and bridging the pores of the 3D-printed mesh. Adapted with permission from references [29] and [92]



mechanical and biological performance. This dual-scale manufacturing enables the creation of dual-functioning patches in which the 3D-printed macrostructure provides temporary mechanical strength to withstand dynamic intra-abdominal forces, while the electrospun micro/nanofibrous layers mimic the extracellular matrix and regulate the biological interface (Fig. 12). Depending on clinical needs, the printed and the electrospun layers can be functionalized to have different roles. Typically, the electrospun layer is engineered to serve as an anti-adhesive barrier facing the visceral side, as demonstrated by Hu et al., who used a SiO_2 -modified electrospun surface to reduce unwanted cellular adhesion [93]. The 3D-printed macrostructure, on the other hand, is commonly made of polymers such as PCL facing the parietal side to provide robust mechanical support, with both studies by Hu et al. reporting tensile strengths safely exceeding the clinical threshold required for abdominal wall repair [29, 93].

The incorporation of bioactive natural polymers into these scaffolds enhances biological performance. Collagen, SP, and κ -carrageenan have been shown to promote endothelial proliferation, neovascularization, and tissue integration while suppressing pathological fibrosis. To control infections, one of the most critical postoperative complications, printed meshes can be loaded with antibiotics to achieve localized and sustained release. This strategy was demonstrated by Deveci et al. and Hu et al., who reported controlled delivery of ciprofloxacin for over 28 days and vancomycin for approximately 10 days, respectively [9, 93]. However, this comparison does not quantitatively demonstrate antimicrobial effectiveness relative to pathogen-specific minimum inhibitory concentrations. In addition, a coaxial approach can be used in both 3D printing and electrospinning to create a core-shell structure that protects sensitive molecules during manufacturing, enables biphasic release, and encapsulates hydrophilic drugs to prevent burst release.

Despite the identified advantages, the reported studies present some limitations, including short in vivo durations and the use of small animals. Since the animal models used cannot replicate exactly the biomechanical environment and healing kinetics found in humans, future studies must evaluate the biopatches in large animal models with closer anatomical similarity over extended periods. An important consideration in hybrid

electrospun/3D-printed structures is the adhesion between layers, which is a common concern in hybrid multilayer systems. Direct electrospinning onto the 3D-printed structure is commonly employed, with interlayer adhesion being promoted by electrostatic attraction and physical entanglement. Adhesion between layers may be further enhanced by using chemically compatible materials, increasing substrate roughness, applying plasma or chemical surface treatments, partially softening the printed surface with a suitable solvent, or introducing a cross-linker as an intermediate layer. Nevertheless, insufficient interfacial adhesion may result in delamination during handling, implantation, or cyclic loading. Quantitative evaluation of interfacial adhesion under clinically relevant loading conditions remains limited in most reported hybrid systems and remains a critical missing requirement in the current literature.

Despite the complementary functions provided by 3D-printed and electrospun layers, hybrid systems introduce additional engineering challenges that are not present to the same extent in the individual technologies. Residual stresses generated during processing, differences in layer thickness and fiber orientation, and mismatched mechanical properties may also affect the overall deformation behavior of the construct. Dense electrospun layers may additionally influence mass transport across the mesh, while sterilization may alter both the individual materials and the integrity of the interface. For biodegradable hybrid systems, differences in degradation rates between layers may further compromise mechanical stability over time.

7 Limitations, challenges, and future perspectives in hernia mesh development

Although this review summarizes the advances achieved in the development of hernia repair meshes, particularly through advanced manufacturing approaches, there are still several limitations and challenges that hinder their application and clinical translation. As mentioned in Sects. 4–6, the main limitation associated with these different production processes lies in the mechanical properties of the developed structures. Although some of the reported studies, particularly those involving extrusion-based 3D printing (Table 7), show values of Young's modulus and tensile strength above the commonly cited reference thresholds, these parameters do not take into account the anisotropic behavior and complex load distribution of the native abdominal wall. This mechanical mismatch is particularly relevant to electrospun structures, which normally present limited load-bearing capacity and may exhibit isotropic behavior [94].

In addition to mechanical limitations, the biological response of hernia meshes remains a critical challenge. Although advanced manufacturing techniques, particularly electrospinning, provide a versatile platform for the use of biocompatible polymers, clinical performance depends on complex interactions at the tissue–material interface. Inadequate integration may result in fibrosis, limited vascularization, persistent inflammatory responses, and foreign body reactions, leading to complications such as chronic pain and adhesion formation [95]. Drug delivery systems have been widely explored to address these challenges; however, as discussed in Sects. 4–6, most of the reported studies still fail to achieve controlled release profiles capable of sustaining long-term therapeutic effectiveness. However, antibacterial activity demonstrated *in vitro*, including zones of inhibition, does not necessarily translate into the prevention of surgical site infection *in vivo*.

Adhesion management represents a further challenge because mesh fixation and visceral protection require opposite surface behaviours, as discussed in Sect. 3.3. While an effective method for fixation to the abdominal wall requires sufficient adhesion, the prevention of interactions with other surrounding tissues, particularly the viscera, demands anti-adhesive properties [96]. If these functions are not spatially controlled, self-adhesive designs may contribute to fibrosis or unwanted tissue attachment. In this way, the development of asymmetric or multilayer meshes, by the use of coatings, different production techniques, or even different materials within the same technique, can be a promising solution. Advanced manufacturing techniques, particularly extrusion-based 3D printing and electrospinning, emerge as a high-potential approach for the development of such spatially

differentiated systems, in which each surface is tailored to a specific function. Moreover, these techniques allow for the development of materials capable of dynamically modulating their adhesive properties over time, which represents a promising direction for improving both fixation and long-term biocompatibility.

Another limitation identified in the reviewed studies is the low technological/clinical readiness level of advanced manufacturing approaches for the development of hernia meshes. As discussed in Sects. 4–6, many of the reported studies are still limited to *in vitro* evaluations, while *in vivo* studies are predominantly performed in small animal models that do not accurately replicate the human abdominal wall. In addition, the lack of process standardization, together with challenges related to reproducibility and scalability, remains a significant barrier to reaching clinical level [97]. These issues are particularly evident in electrospun materials, where the high sensitivity of the process to several parameters can lead to variability in the final structure and performance. The lack of long-term validation and clinical studies is therefore critical, but this variability may also be viewed as an opportunity for future development. The versatility of additive manufacturing techniques offers the potential for patient-matched meshes, enabling adaptation to defect variability. This capacity for customization represents a promising direction for overcoming the current limitations [98].

Another challenge is the trade-off between permanent and biodegradable materials in hernia meshes. As discussed in Sect. 3, most commercially available meshes are based on non-absorbable polymers that remain in the body indefinitely, providing long-term mechanical support but also posing a persistent risk of late-stage complications. On the other hand, resorbable meshes produced from biodegradable materials offer the advantage of gradual resorption in the body. However, their mechanical performance may be insufficient to provide support throughout the entire healing process. Ideally, a mesh should provide mechanical support throughout tissue healing and remodeling, and then undergo a controlled and gradual degradation [26]. This temporal mismatch between the loss of implant mechanical integrity and the restoration of sufficient native tissue strength therefore remains a significant design issue that advanced manufacturing techniques may address through the development of hybrid and multi-scale structures, combining different materials with different degradation profiles. The design of such systems must also consider the toxicity of degradation by-products and their influence on the abdominal environment, but a controlled degradation behavior presents a promising approach to next-generation hernia repair meshes.

Beyond material and biological performance, the clinical translation of these hybrid hernia meshes requires a quality-control framework covering the complete manufacturing workflow. In accordance with regulatory guidance, this framework should include controls over digital design, raw material specifications, printing and post-processing parameters, and final product testing [99, 100]. Sterilization methods must be validated according to their effects on the mechanical and biological properties of the mesh. A risk-management framework should evaluate the biological safety through the analysis of chemical constituents, contaminants and residues, degradation products, and microbial contamination [101, 102]. Preclinical studies should use standardized implantation sites, follow-up periods, and representative animal models. In the case of patient-matched devices, these should not automatically be classified as custom-made devices, but rather evaluated within a validated and reproducible process and remain subject to the usual pre- and post-market requirements, according to the International Medical Device Regulators Forum [100].

Economic aspects represent additional challenges for the widespread adoption of advanced manufacturing strategies as generalized procedures in hernia mesh development. Despite the design flexibility and customization potential of these techniques, their cost-effectiveness remains uncertain, particularly for electrospinning, which requires specialized equipment and expertise. Together with the quality-control and regulatory requirements discussed above, these developments may support the clinical translation of next-generation hernia meshes. Nevertheless, these challenges also open new opportunities for innovation, and advances in process and design standardization and scalable manufacturing strategies may contribute to reducing costs and improving reproducibility. At the same time, updated regulatory pathways tailored to additive manufacturing could facilitate clinical translation. In this context, the combination of advances in these techniques with the adaptation of the clinical and regulatory framework may help unlock the full potential of next-generation hernia treatment.

Based on the limitations identified throughout this review, translation towards clinical application should progressively address reproducible material and processing specifications, standardized mechanical and biological evaluation, sterilization and storage stability, long-term mechanical and degradation performance, validation in clinically relevant *in vivo* models, manufacturing scale-up and quality control, and, ultimately, regulatory and economic requirements. Although these aspects are strongly interconnected rather than strictly sequential, they provide a practical framework for guiding future development towards clinical translation.

8 Conclusion

In summary, advanced manufacturing techniques show considerable promise for the development of next-generation hernia meshes. As highlighted in this review, traditional commercial meshes, while effective at reducing hernia recurrence, are frequently associated with severe postoperative complications, including chronic inflammation, visceral adhesions, and mesh-related infections. To overcome these clinical challenges, recent literature demonstrates a potential complementarity between 3D printing and electrospinning. This dual-scale manufacturing approach enables the creation of highly customized dual-functioning meshes. Within these hybrid patches, the 3D-printed macrostructure provides the temporary mechanical support required to withstand dynamic intra-abdominal forces. Simultaneously, the electrospun micro/nanofibrous layers mimic the native ECM, actively regulating the biological interface to promote cellular integration. The incorporation of bioactive natural polymers into these structures has been demonstrated to significantly promote endothelial proliferation, neovascularization, and tissue integration, while also reducing bacterial colonization and fibrotic tissue formation.

Significant progress has also been made in postoperative infection control and immunomodulation. Instead of relying solely on systemic therapies, these meshes are increasingly designed as localized drug delivery systems. Using advanced fabrication techniques such as coaxial electrospinning and 3D printing, researchers can develop core-shell structures that protect sensitive bioactive molecules, minimize initial burst release, and enable sustained long-term release. Despite these significant design improvements, applying these technologies in clinical practice faces critical limitations. Most current research remains limited to *in vitro* proof-of-concept studies or short-term *in vivo* tests using small animal models, which do not fully replicate the biomechanical environment and healing processes in humans. Another challenge is the transition from laboratory fabrication to industrial production.

In conclusion, hybrid manufacturing techniques have significantly expanded the design possibilities for hernia repair meshes, allowing for personalized, bioactive implants that actively support tissue regeneration while reducing complications. The most critical unresolved gap is the limited translational validation of the several hernia meshes produced by additive manufacturing under clinically relevant conditions. Although the combined use of 3D printing and electrospinning provides a versatile platform for addressing the complex challenges of hernia repair, their long-term behavior under abdominal loading and progressive degradation remains insufficiently understood. In this context, future research should prioritize standardized testing protocols combining mechanical assessment with long-term degradation and biocompatibility. Addressing this will be essential for identifying the potential of the proof-of-concept designs for clinical translation.

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Author contributions A.C.N. and M.C.F. performed the research and wrote the main manuscript text. R.P.C.L.S., R.F. and D.P.F. conceptualized and supervised the work, and reviewed the full manuscript. All authors have reviewed the manuscript.

Funding The authors acknowledge the financial support from the integrated project GIATEX—Gestão Inteligente da Água na ITV (Investment RE-C05-i01.01—Mobilizing Agendas/Alliances for Business Innovation), promoted by the Recovery and Resilience Plan (RRP), Next Generation EU, for the period 2022–2025. This work was also funded by national funds

through FCT – Fundação para a Ciência e a Tecnologia, I.P. and Programa de Recuperação e Resiliência (PRR) through NextGenerationEU from the European Union, under the Strategic Projects UID/00264/2025 and UID/PRR/00264/2025 of the 2C2T – Centro de Ciência e Tecnologia Têxtil (<https://doi.org/10.54499/UID/00264/2025> and <https://doi.org/10.54499/UID/PRR/00264/2025>).

Data availability No datasets were generated or analysed during the current study.

Declarations

Ethics approval and consent to participate Not applicable.

Consent for publication Not applicable.

Competing interests The authors declare no competing interests.

Clinical trial number Not applicable.

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