Table 1. Classification of commercially available first generation surgical meshes [38].

Product (Manufacturer)	Material	Pore Size (mm)	Absorbable	Weight (g/m^2)	Filament	Mechanical Properties	Advantages and Disadvantages
Vicryl (Ethicon)	Polyglactin	0.4	Yes, fully (60–90 days)	56	Multifilament	Tensile strength of 78.2 ± 10.5 N/cm in longitudinal direction and 45.5 ± 13.5 N/cm in transverse direction.	Eliminates the risk of infectious disease transmission. Usually results in hernia recurrence after complete absorption
Dexon (Syneture)	Polyglycolic acid	0.75	Yes, fully (60–90 days)	56	Multifilament	N.A.	Adhesions fade as the mesh is absorbed. It is controversial whether the fibrous ingrowth into the prosthesis is sufficient to accomplish a permanent repair.
Sefil (B-Baun)	Polyglycolic acid	0.75	Yes, fully (60–90 days)	56	Multifilament	N.A.	High anatomic adaptability and low risk of late secondary infection. Retain 50% of its strength for 20 days.
Marlex (BARD)	PP	0.8	No	80–100	Monofilament	Tensile strength of 58.8 N/cm	High tensile strength. Evokes a chronic inflammatory reaction.
3D Max (BARD)	PP	0.8	No	80–100	Monofilament	Tensile strength of 124.7 N/cm	Anatomically designed. Reduced patient pain. Adhesions risk.
Polysoft (BARD)	PP	0.8	No	80–100	Multifilament	Burst strength of 558 N and a stiffness of 52.9 N/cm	Low infection risk. Not used in extraperitoneal spaces as produce dense adhesions *.
Prolene (Ethicon)	PP	0.8	No	80–100	Monofilament	Tensile strength of 156.5 N/cm	Facilitates fibrovascular ingrowth, infection resistance and improve compliance. Adhesions risk.
Surgipro (Autosuture)	PP	0.8	No	80–100	Multifilament	Tensile strength of 41.8 N/cm in longitudinal direction and 52.9 N/cm in transverse direction	High tensile strength, ease of handling and position and retains properties in vivo. Difficult complete wound healing caused by mesh structure.
Prolite (Atrium)	PP	0.8	No	80–100	Monofilament	Tensile strength of 138 N/cm	Monofilaments aligned in parallel spaced angles to maximizing material flexibility in two dimensions and a smooth and very uniform open architecture. Adhesions risk.
Trelex (Meadox)	PP	0.8	No	80–100	Multifilament	N.A.	*
Atrium (Atrium)	PP	0.8	No	80–100	Monofilament	Tensile strength of 56.2 N/cm	High tolerance to infection. Adhesions risk.

Table 1. *Cont.*

Product (Manufacturer)	Material	Pore Size (mm)	Absorbable	Weight (g/m^2)	Filament	Mechanical Properties	Advantages and Disadvantages
Premilene (B-Braun)	PP	0.8	No	80–100	Monofilament	Tensile strength of 41.4 N/cm in longitudinal direction and 36.5 N/cm in transverse direction	Mesh adaptation to the longitudinal and latitudinal axes of the connective tissue where is used for the reinforcement, rapid healing and tissue penetration. Adhesions risk.
Serapren (smooth)	PP	0.8	No	80–100	Multifilament	N.A.	*
Parietene (Covidien)	PP	0.8	No	80–100	Multifilament	Tensile strength of 38.9 $\pm$ 5.2 N/cm in longitudinal direction and 26.6 $\pm$ 4.2 N/cm in transverse direction	*
Prolene Light (Covidien)	PP	1.0–3.6	No	36–48	Monofilament	Tensile strength of 20 N/cm	Greater flexibility. Not used in intraperitoneal spaces as produce dense adhesions.
Optilene (B-Baun)	PP	1.0–3.6	No	36–48	Monofilament	Tensile strength of 58 N/cm	Soft, thin and pliable. Ideal for inguinal hernia repair to reduce chronic pain. Not used in extraperitoneal spaces as produce dense adhesions.
Mersilene (Ethicon)	POL	1.0–2.0	No	40	Multifilament	Tensile strength of 19 N/cm	Low infection risk. Evokes an aggressive macrophage and giant cell rich inflammatory reaction, followed by a dense fibrous ingrowth.
Goretex (Gore)	e-PTFE	0.003	No	Heavyweight	Multifilament	Minimum tensile strength of 16 N/cm	Smooth and strong. Evokes a chronic inflammatory reaction.

PP: Polypropylene. POL: Polyester. e-PTFE: Expanded polytetrafluoroethylene. N.A, Information not available in literature. * Duplicated properties.

Table 2. Classification of commercially available second generation surgical meshes [38].

Product (Manufacturer)	Material	Pore Size (mm)	Absorbable	Weight (g/m^2)	Filament	Mechanical Properties	Advantages and Disadvantages
Vypro, Vypro II (Ethicon)	PP/polyglactin 910	>3	Partially (42 days)	25 & 30	Multifilament	Tensile strength of 16 N/cm	Significantly decreased rates of chronic pain. Higher rate of hernia recurrence.
Gore-Tex Dual Mesh Dual Mesh Plus (Gore)	e-PTFE	0.003–0.022	No	Heavyweight	Multifilament	Minimum tensile strength of 16 N/cm (Gore-Tex Dual Mesh) and 157.7 N/cm (Dual Mesh Plus)	Promotes host tissue growth and reduces tissue attachment. Infection risk.
Parietex (Covidien)	POL/collagen	>3	Partially (20 days)	75	Multifilament	Elasticity of 3.5 at 16 N	Short-term benefit for anti-adhesion property. Greater infection rate (57%).
Composix EX Dulex (BARD)	PP/e-PTFE	0.8	No	Lightweight	Monofilament	N.A.	Minimizes adhesions and provides optimal tissue ingrowth. Infection risk.
Proceed (Ethicon)	PP/cellulose	Large	Partially (<30 days)	45	Monofilament	Tensile strength of 56.6 N/cm	Low rates of hernia recurrence (3.7%). Risk of formation of visceral adhesions.
DynaMesh IPOM (FEG Textiltechnik)	PP/PVDF	1–2	Partially	60	Monofilament	Tensile strength of 11.1 ± 6.4 N/cm in longitudinal direction and 46.9 ± 9.7 N/cm in transverse direction	Minimal foreign body reaction. Adhesions risk.
Sepramesh (Genzyme)	PP/sodium	1–2	Partially (<30 days)	102	Monofilament	N.A.	Reduces adhesions and the optimal tissue ingrowth is promoted. Sticky consistency difficult the surgeon manipulation.
Ultrapro (Ethicon)	PP/PGC-25	>3	Partially (<140 days)	28	Monofilament	Tensile strength of 55 N/cm	Reduced inflammatory response. Adhesions risk.
Ti-Mesh (GfE)	PP/titanium	>1	No	16 & 35	Monofilament	Tensile strength of 12 N/cm (mesh of 16 g/m^2) and 47 N/cm (mesh of 35 g/m^2)	Reduced inflammatory response. Low tensile strength.
C-Qur (Atrium)	PP/omega 3	>1	Partially (120 days)	50	Monofilament	Ball burst strength of 170 ± 20.1 N	Short-term benefit for anti-adhesion property. No significant difference for adhesion grade or amount relative to other meshes.

PP: Polypropylene. e-PTFE: Expanded polytetrafluoroethylene. POL: Polyester. PVDF: Polyvinylidene fluoride. PGC-25: poliglecaprone 25. N.A, Information not available in literature.

Table 3. Classification of commercially available third generation surgical meshes [38].

Product (Manufacturer)	Material	Tensile Strength (MPa)	Advantages	Disadvantages
Surgisis (Cook)	Porcine (small intestine submucosa)	4	No refrigeration is required. Long history of safety data.	Requires hydration. Susceptible to collagenases.
FlexHD (J&J)	Human (acellular dermis)	10	No refrigeration or rehydration is required.	N.A.
AlloMax (Davol)	Human (acellular dermis)	23	No refrigeration or rehydration is required. Available in large sizes.	Hydration required.
CollaMend (Davol)	Porcine/Bovine (xenogenic acellular dermis)	11	No refrigeration or rehydration is required. Available in large sizes.	N.A.
Strattice (LifeCell)	Porcine/Bovine (xenogenic acellular dermis)	18	Available in large sheets.	Limited long-term follow up.
Permacol (Covidien)	Porcine/Bovine (xenogenic acellular dermis)	39	No refrigeration or rehydration is required. Available in large sizes.	N.A.
XenMatrix (Davol)	Porcine/Bovine (xenogenic acellular dermis)	14	Available in large sheets.	Limited long-term follow up.

N.A. Information not available in literature.

3.7. Manufacturing Processes for Surgical Meshes

Surgical meshes are produced from different synthetic materials and in different mesh structures, the knitted structure being the most common [44]. Surgical filaments are mainly manufactured by extrusion processes and then knitted accordingly. As mentioned, meshes are typically manufactured from PL, PP, PTFE, e-PTFE, PVDF and composite materials (e-PTFE/PP) [45]. The knitting pattern can be significantly altered resulting in a broad range of properties. Thickness, pore size, tensile strength, flexural rigidity, and surface texture are highly dependent upon the knitting pattern; the resultant interplay among these characteristics imparts different performance [44]. These characteristics, besides altering the biocompatibility of the mesh given its affinity to cells, also dictate the mechanical properties of the mesh such as rigidity and deformation. Knitted meshes are a subset of the non-woven mesh configuration. However, there is much more order and consistency with pore size using a knitted design [46]. Knitting, by definition, is the construction of a fabric or cloth from the interlocking of threads through the formation of loops. Recent studies have been focused on treating the surgical mesh as a high-tech textile rather than as a prosthesis [44].

3.7.1. The Extrusion Process

Melt extrusion is the least expensive and simplest form of fiber extrusion [47]. This process consists of melting the polymer pellets through a combination of applied heat and friction. The molten polymer is then forced under high pressure through a small orifice or a "shower head" spinneret. The molten polymer flows out of the spinneret and freezes into a solid fiber, which is then typically reheated and drawn numerous times to further align the molecules and hence strengthen the fiber [48].

Most of the surgical meshes are made from filaments initially developed to be used for surgical sutures. Surgical sutures are made from polymers like PP [49], PL [50], e-PTFE [51] or PVDF [52] monofilaments and have been successfully used by the medical profession for decades. Filaments used for surgical sutures have to possess several characteristics such as [53]:

1. Ability to attach to needles by the usual procedure.
2. Capability to be sterilized using ethylene oxide or ultraviolet radiation.
3. Ability to pass easily through tissue.
4. Ability to resist breakdown without developing an infection.
5. Possess minimal reaction with tissue.
6. Maintain its in vivo tensile strength over extended periods.

The most commonly used systems in the knitting manufacturing process are the Tricot [60] and Raschel knitting machines [61], which are used to create warp or weft knitting structures [62]. Warp knitted meshes are the most popular system used to repair hernia defects, and are manufactured using the Raschel machine with a basic configuration consisting of two bars where latch-type needles are collectively mounted (running the full knitting width of the machine) and guide bars to hold yarn beams individually. The needle bars follow up and down movements, while the guide bars move back and forth across the needles of each bar to form continuous loops. The warp knit fabric design and lapping sequence is controlled by the shagging or traverse motion of the guide bars [63].

In principle, the Tricot knitting machine is very similar to the Raschel knitting; the only difference is the use of spring beard or compound needles instead of the latch needles used in the Raschel knitting machine. In addition, Tricot sinkers not only performed the function of holding down the loops whilst the needles rise as Raschel sinkers, but also support the fabric loops. The small angle of fabric take-away and the type of knitting action in Tricots creates a gentle and lower tension on the knitted fabric, ideal for high-speed production of fine gauge [64].

A double Raschel warp knitting machine (DR 16 EEC/EAC) has 16 guide bars and enables the production of textiles with different yarn materials and counts. The machine is equipped with two different gauges, E18 and E30. This system allows the design of a mesh configuration that could be adjusted to match given design parameters such as size, shape, Young modulus, and porosity [65]. The ultimate mechanical properties of the meshes are determined by the intrinsic properties of the filaments and the final configuration of the knitted fabrics.

4. Future Perspectives

Despite the clinical success and vast body of knowledge that has been gained regarding manufacturing of surgical meshes, material properties, and surgical procedures, it is obvious that the ideal mesh has not been developed. It is well known that meshes still suffer from contraction and/or infection after implantation [66]. Furthermore, adhesions between the visceral side of the mesh and adjacent organs still occur. These complications may have serious consequences, such as chronic pain, intestinal obstruction, bowel erosion, or hernia recurrence. All of these problems have opened a great number of opportunities to create a new generation of surgical meshes [67]. This new generation will have to show a better integration with the tissue of the abdominal wall, but no adhesions on the visceral side. Based on the ideas of van't Riet [68], Ebersole [69] and Xu [70], new alternatives rely broadly on surface mesh modification by novel coatings to existent meshes and/or integration of nanofiber based systems.

4.1. Coatings

A variety of biocompatible and biodegradable natural and synthetic polymers are being investigated. Extensive research focuses in the development of a bi-layer composite hernia mesh in order to minimize the risk of infections and reduce adhesions on the visceral side [71,72]. Materials that had been studied are: Polylactic acid (PLLA) [20], oxygenated regenerated cellulose (ORC) [67], n-vinyl pyrrolydone (NVP) and n-butylmethacrylate (BMA) [67], polyglycolic acid (PGA) [73], carboxymethylcellulose (SCMC) [74], omega-3 fatty acid [75], messenchymal stem cells (RMSC) [76], human dermal (HDF) and rat kidney fibroblasts (RKF) [76], collagen [77–79], chitosan [80], nanocrystalline silver particles (NCSP) [81] and titanium [82,83]. Table 5 shows some of the properties that have made these materials attractive as active ingredients in surgical meshes [71,80,84–86].

Most of the recently published literature still presents PP surgical meshes as the "gold standard" though with surface modifications made with materials mentioned in Table 5. Studies have primarily concentrated on: thickness and concentration of the materials used in the coating to be in contact with the visceral and/or abdominal side (Ex: 95% of oxidized collagen and 5% of chitosan) [26] and surface density (measured in g/m^2). The following Table 6 presents a summary of the obtained results based on the inflammatory response and percentage of adhesion.

Table 5. Material properties of surgical mesh coatings.

PLLA/PGA	ORC/SCMC	NVP/BMA	Omega-3 Fatty Acid	RMSC/HDF/RKF	Collagen/Chitosan	NCSP	Titanium
Variable degradation rate	Reduce mesh adhesions	Reduce mesh adhesions	Minimal risk of mesh contraction	Affinity towards fibroblasts	Weak tensile properties	Anti-inflammatory	Provides mechanical integrity
Hydrophilicity	Absorbable	Hydrophilicity	Absorbable	Favourable cell adhesion	Negligible effect on biomechanical properties	Antimicrobial	Non-absorbable

PLLA: Polylactic acid. PGA: Polyglycolic acid. ORC: Oxygenated regenerated cellulose. SCMC: Carboxymethylcellulose. NVP: N-vinyl pyrrolydone. BMA: N-butylmethacrylate. RMSC: Messenchymal stem cells. HDF: Human dermal. RKF: Rat kidney fibroblasts. NCSP: Nanocrystalline silver particles.

Table 6. Examples of surgical mesh coating parameters.

Reference	Analyzed Parameter	
	Material	**Surface Density**
Pascual et al. [86]	Oxidized collagen Chitosan	Oxidized collagen 95%/ Chitosan 5%
Ciechańska et al. [71]	MBC	6.7 g/m^2 (one side) 5.31 g/m^2 (two sides)
Cohen et al. [81]	NCSP	310 g/m^2 640 g/m^2 1130 g/m^2
Niekraszewics et al. [85]	Chitosan	20 g/m^2 (one side) 20 g/m^2 (two sides)

MBC: Modified bacterial cellulose. NCSP: Nanocrystalline silver particles.

In general, the new composite meshes show highly improved performance regarding peritoneal regeneration and visceral adhesion [84]. These studies have developed composite surgical meshes with high potential for adoption. Further studies with a focus on long-term adhesion and structural performance will complement obtained results.

4.2. Nanofibers

Nanofiber systems made from a large variety of materials have been explored extensively in the last decade. Scaffolds for tissue regeneration are strongly deemed as a potential application of these systems [87]. Mimicking the extracellular matrix (ECM) is vital to control cell behavior, such as adhesion, proliferation, migration, and differentiation. Tissue Engineering (TE) has been extensively explored to provide answers associated with current problems encountered in the interaction of the surgical meshes with the human body. One of the challenges of TE is to mimic the natural extracellular matrix (ECM) of the abdominal wall to promote an efficient integration. Researchers are actively exploring the implementation of nanofiber systems to effectively mimic the ECM [88–90].

Nanofibrous structures present several advantages, such as high specific surface area for cell attachment, higher microporous structure and a 3D micro environment for cell–cell and cell–biomaterial contact, these being associated with unique physical and mechanical properties. These structures when compared with commercial surgical meshes possess higher porosity and smaller pore size. These properties make nanofiber systems suitable for biomaterials used in wound care, drug delivery, and scaffolds for tissue regeneration [20,44,91].

Scaffolds for tissue engineering must possess a porous structure able to facilitate cell migration, a balance between surface hydrophilicity and hydrophobicity for cell attachment, mechanical properties comparable to natural tissue, and biocompatibility. Studies have shown that the abovementioned characteristics are also highly influenced by average diameter of the fibers and pore size. Effective cell attachment and proliferation has been observed in fiber systems with average diameters smaller than 1 µm and average pore size of 14 µm [92]. In commercially available meshes, even when it has been shown that cells are able to proliferate in micrometer/macrometer regimes, the cells in fact have difficulty attaching and proliferating. Cells are seen around the fibers whereas, on nanofiber based meshes, the cells attach to the fibers and quickly proliferate while making strong contact with underlying nanofibers, therefore promoting interlayer growth.

The application of nanofiber systems has been hampered due to its poor mechanical properties and nanofiber availability. Most of the available studies have focused on nanofibers prepared through solution processes. The properties of the developed fibers can be controlled by different parameters such as utilized solvent, concentration of polymer, processing methods, and ambient conditions. For example, in the case of nanofibers made of polypropylene (one of the highly used polymers for commercially available surgical meshes), decahydronaphthalene (decalin) and cyclohexane have been

used as preferred solvents. Polypropylene nanofibers prepared with cyclohexane exhibited a rougher surface when compared to the fibers prepared with decalin, suggesting that the surface morphology of the nanofibers depend on the boiling point of each solvent [93]. When stress–strain behaviors of the nanofibers are investigated, a tensile strength of 61.4 ± 1.5 MPa with 35.2% ± 1.7% of strain, and a Young modulus of 174.6 ± 1.7 MPa was obtained for the decalin based nanofibers, whilst the cyclohexane nanofibers exhibit a tensile strength of 18.2 ± 1.1 MPa with 46.7% ± 1.2% of elongation and a Young modulus of 39.1 ± 1.4 MPa [94]. The abovementioned results were obtained from bundles of nanofibers rather than individual fibers, these properties are strongly dependent on fiber orientation within the tested sample, bonding between fibers, and slip of one fiber over another [94].

Regarding nanofiber availability, there are several methods to prepare nanofiber systems. These methods include wet chemistry, Electrospinning (ES) [95] and Forcespinning® (FS) [96] techniques. Most of the available literature has used ES processes; these studies have proven the potential of these nanofiber systems towards solving many of the challenges encountered in TE. ES processes have been limited to laboratory-based research given the challenges associated with increasing yield and opportunity to work with melt based systems. FS, a technique that has been recently introduced is based on developing nanofibers through the application of centrifugal forces. The method has been proven effective to produce yields that could satisfy industry requirements (i.e., several hundred meters per minute) as well as to produce nanofibers from melt based systems therefore removing the requirement of a solvent and subsequently the potential contamination of the materials with toxic organic solvents, and cost associated with the solvent itself and solvent recovery procedures. Other scaffolds had been produced by 3D printing procedures. Such biomimetic scaffolds are promising techniques as they could allow precise control over the geometry and microstructure [46,97].

Table 7 presents a summary of recently published work regarding the manufacture of nanofiber based surgical meshes.

Table 7. Nanofiber based surgical meshes.

Nanofiber Material	Manufacturing Process	Diameter (nm)	Tensile Strength (MPa)	Advantages and Disadvantages	Reference
Poly-ε-caprolactone (PCL)	Electrospinning	1280 ± 330	3.11 ± 1.09	Better adhesion, growth, metabolic activity, proliferation and viability of 3T3 Fibroblasts. Lack of in vivo testing.	[87,98]
Polydioxanone (PDO)	Electrospinning	860 ± 420	3.76 ± 0.49	Bioresorbable polymer. Reduction of long-term foreign body response (LTFBR). No fulfill the mechanical requirements.	[99]
Polylactide-Co-Glycolide (PLGA 8218)	Electrospinning	3280 ± 570	6.47 ± 0.41	Exceed the minimum mechanical requirements for hernia repair applications. Bioresorbable polymer. Reduction of LTFBR. Lack of in vivo testing.	
PLLA	Electrospinning	1480 ± 670	3.59 ± 0.25	In vivo advantages. Exceed the minimum mechanical requirements for hernia repair applications. Lack of in vivo testing.	
Polyurethane (PU)	Electrospinning	890 ± 330	18.9 ± 5.9	Elastic deformation.	
PET	Electrospinning	710 ± 280	3.17 ± 0.23	Adequate mechanical attributes. No evidence of intestinal adhesions. Trigger of a large foreign body reaction.	[100]
PET/Chitosan	Electrospinning	3010 ± 720	2.89 ± 0.27	Adequate mechanical attributes. No evidence of intestinal adhesions. Trigger of a large foreign body reaction.	
PCL/Collagen	Electrospinning	1000	2.13 ± 0.36	Biological and biomechanical stable, support skeletal muscle cell ingrowth and neo-tissue formation	[101]

PCL: Poly-ε-caprolactone. PDO: Polydioxanone. PLGA 8218: Polylactide-Co-Glycolide. PU: Polyurethane. PET: Polyethylene terephthalate.

Nanofiber systems are certainly showing a strong potential to be used in the next generation of surgical meshes, the increased availability (FS process) will certainly promote the development of practical applications. Nanofiber developed through the FS system have shown promising results regarding adhesion, growth, metabolic activity, proliferation, and viability of 3T3 cells [70,102]. It is expected that these systems will be used in combination with existent commercial meshes to satisfy other requirements such as mechanical strength needed to bear the intra-abdominal pressure exerted by human body and implantation requirements to mention some. Future studies in this area will include the effect of nanofiber morphology, mesh design (i.e., uniaxial aligned, radially aligned, orthogonally patterned) needed to improve structural properties, and in vivo testing.

In summary, this review synergistically complements recent reviews made in this important area. Table 8 presents a comparative table with recent published reviews [38,103–106]. Besides having in common the history and present scenario, this review also presents information regarding manufacturing methods (manufacturing of these meshes has a strong influence in the medical results, therefore the ultimate functionality will be strongly dependent upon the manufacturing method) and future perspectives.

Table 8. Aspects related to hernia meshes compared in recently published reviews.

	Baylon et al. (This Review)	**Brown et al. [38]**	**Sanbhal et al. [103]**	**Guillaume et al. [104]**	**Todros et al. [105]**	**Todros et al. [106]**
Introduction	√	√	√	√	√	√
History	√	√	-	-	-	-
Present Scenario	√	√	√	√	√	√
Properties Discussed	Elasticity/tensile strength Pore Size Weight (density) Constitution Material absorption	Tensile strength Pore Size Weight Reactivity/Biocompatibility Elasticity Constitution Shrinkage Complications	Weight Pore Shape, size/porosity Mesh elasticity/strength	Properties discussed for particular meshes, varies from the type of mesh being discussed.	Pore size Density thickness	Biomechanical properties Uniaxial tensile testing Biaxial tensile testing Ball burst testing
Surgical Mesh	√	√	√	√	√	√
Manufacturing Processes	> 2 processes considered	-	-	-	-	-
Future Perspectives	2 perspectives considered	-	√	√	-	-
Comments	Comparison of meshes divided by generations: First generation (18 meshes), second generation, (10 meshes), third generation (7 meshes)	Comparison of meshes divided by constitution, Multi (3 meshes), multifilament and monofilament (13 meshes), and foil (1 mesh). Biomaterial meshes (10 meshes)	Comparison between synthetic meshes (15 meshes) Comparison between composite meshes (12 meshes)	Meshes divided by Biologically Derived Matrices, Biodegradable synthetic structures, Anti-inflammatory mesh, Meshes with enhanced cytocompatibility, Anti-adhesive Mesh, Antibacterial meshes. Review also discusses mesh fixation, self-expanding systems, post-implantation visible mesh, cell coated meshes, and growth factor loaded meshes.	Comparison between synthetic surgical meshes: HWPP (5 meshes), LWPP (6 meshes), PET (1 mesh), ePTFE (1 mesh), PVDF (1 mesh) Comparison between Multilayered meshes (10 meshes)	Comparison between synthetic surgical meshes: HWPP (5 meshes), LWPP (3 meshes), PET (1 mesh), ePTFE (1 mesh), PVDF (1 mesh). Comparison between Multilayered Meshes (10 meshes)
Total meshes compared	35	27	27	-	24	21

5. Conclusions

Surgical meshes have become the system of choice for hernia repair. Even though it is not the optimum method, so far it is the one that has shown a lower rate of recurrence. Currently, there are more than 70 types of meshes commercially available. These are constructed from synthetic materials (absorbable, non-absorbable, or a combination of both) and animal tissue. Despite reducing rates of recurrence, hernia repair with surgical meshes still faces adverse effects such as infection, adhesion, and bowel obstruction. Most of these drawbacks are related to the chemical and structural nature of the mesh itself.

An optimum integration with the abdominal wall and negligible adhesion on the visceral side are the most important after sought features for the "ideal" mesh. A surgical mesh will trigger one of three different responses from the body: it may be integrated, encapsulated or degraded. In order to have a minimal inflammatory response to better integrate it to the body, it is highly important to improve biocompatibility.

To overcome this obstacle, researchers are actively exploring methods to improve biocompatibility, with the goal of developing a mesh that can be effectively incorporated with minimal inflammation and/or infection. Nanofibers have been recently considered as a strong potential intermediary structure to be used as a coating, given their ultralightweight quality, which could contribute to minimize the inflammatory response from the body and given its functional porosity, which could promote cell adhesion and proliferation.

Acknowledgments: The authors gratefully acknowledge support received by the National Science Foundation Partnership for Research and Education in Materials (PREM) award under Grant No. DMR-1523577: The University of Texas Rio Grande Valley–University of Minnesota Partnership for Fostering Innovation by Bridging Excellence in Research and Student Success. This work was also funded by Tecnológico de Monterrey—Campus Monterrey, through the Research group of Nanotechnology and Devices Design. Additional support was provided by Consejo Nacional de Ciencia y Tecnología (CONACYT), Project Number 242269, Mexico.

Author Contributions: As a review article, all authors contributed to the writing, editing and revision of the manuscript. All authors contributed equally to the development of this review article.

Conflicts of Interest: The authors declare no conflict of interest.

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