REVIEW

Surgical mesh for ventral incisional hernia repairs: Understanding mesh design

Ali Rastegarpour MD[1], Michael Cheung MD[1], Madhurima Vardhan MS[2], Mohamed M Ibrahim MD[1], Charles E Butler MD FACS[3], Howard Levinson MD[1]

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Surgical mesh has become an indispensable tool in hernia repair to improve outcomes and reduce costs; however, efforts are constantly being undertaken in mesh development to overcome postoperative complications. Common complications include infection, pain, adhesions, mesh extrusion and hernia recurrence. Reducing the complications of mesh implantation is of utmost importance given that hernias occur in hundreds of thousands of patients per year in the United States. In the present review, the authors present the different types of hernia meshes, discuss the key properties of mesh design, and demonstrate how each design element affects performance and complications. The present article will provide a basis for surgeons to understand which mesh to choose for patient care and why, and will explain the important technological aspects that will continue to evolve over the ensuing years.

Key Words: *Hernia; Surgical mesh; Tissue engineering; Ventral*

Le treillis chirurgical pour réparer l'incision des hernies ventrales : comprendre la conception du treillis

Le treillis chirurgical est devenu indispensable pour réparer les hernies, car il améliore les résultats et réduit les coûts. Cependant, les treillis sont en constant développement afin de vaincre les complications postopératoires. Parmi les complications courantes, soulignons l'infection, la douleur, les adhérences, l'extrusion du treillis et la récurrence des hernies. Il est essentiel de réduire les complications liées à l'implantation des treillis, car des centaines de milliers de patients souffrent de hernies chaque année aux États-Unis. Dans la présente analyse, les auteurs présentent les divers types de treillis pour hernie, en exposent les principales propriétés et démontrent l'effet de chaque élément de conception sur le rendement et les complications. Le présent article aidera les chirurgiens à choisir le treillis pour leurs patients et exposera les aspects technologiques importants qui continueront d'évoluer au cours des prochaines années.

Incisional hernia is the most common complication of laparotomy that requires reoperation. Recent figures cite an overall incidence of nearly 10% (1). Considering that two million laparotomies are performed annually in the United States (2), there will be an estimated 200,000 patients requiring incisional hernia repair each year (3). For stoma site hernias, the incidence of hernia formation may be as high as 30% and, when surgical site infections occur, the incidence is believed to double (4,5). The costs of incisional hernia repair surgeries are staggering. Poulose et al (6) calculated an average cost of USD$15,899 for each in-patient operation in the United States in 2006, which amounts to an estimated $3.2 billion per year. Bower and Roth(7) were quick to point out that this is likely an underestimation of total costs, because the study by Poulose et al (6) did not account for physician fees and societal costs, such as absence from work, and excluded Veterans Affairs (VA) system costs.

Nonincisional hernias share many aspects of their pathophysiology and management with incisional hernias. Collectively, the repair of nonincisional abdominal wall hernias form the most common group of major operations performed by general surgeons, with more than one million procedures annually in the United States (8). These hernias demonstrate a prevalence of 1.7% in the general population, rising up to 4% in individuals >45 years of age. Inguinal hernia, which accounts for 75% of these occurrences, holds a lifetime risk of 27% in men (9).

Prosthetic meshes are widely applied to reduce hernia recurrence rates. The 10-year incisional hernia recurrence rate is reported to be 63% for traditional suture repair without mesh and 32% for repairs using prosthetic mesh (10). While meshes are obviously beneficial, they remain associated with several serious complications including hernia recurrence, infection (11), chronic pain (12) and adhesions (13). As such, many hurdles remain to be overcome with new hernia mesh designs. The present article reviews the different classes of hernia meshes and principles of tissue engineering as applied to mesh development, and explains how current complications associated with surgical mesh are being addressed with different mesh designs.

VENTRAL INCISIONAL HERNIA

During closure of a laparotomy, the linea alba is reapproximated and the rectus muscles are returned to midline. The integrity of the repair is dependent on suture fixation until the load-bearing properties of the scar become equal to or surpass that of the suture. The fundamental pathophysiology of ventral incisional hernia is lateral migration of the rectus muscle with loss of function commonly referred to as 'loss of domain' (14). Mesh has become standard for repair of incisional hernias because it mitigates loss of domain and helps maintain the rectus muscles in the midline where they function best. The impact of mesh was clearly demonstrated in a multicentre randomized study published in the *New England Journal of Medicine* (15). Luijendijk et al (15) reported that patients undergoing standard suture repair experienced a recurrence rate of nearly double that of patients with mesh repair. Similarly, in a recent meta-analysis published in the *Journal of the American Medical Association, Surgery*, patients undergoing suture repair experienced a nearly threefold increase in hernia recurrence rates when compared with patients who underwent mesh repair (16). As such, the current recommendations set forth by the Ventral Hernia Working Group include the use of mesh to reinforce all ventral hernia repairs (17). Further recommendations include centralization and reapproximation of the paired rectus muscles. In selected instances, when the rectus muscles are splayed apart and cannot easily come

[1]*Division of Plastic and Reconstructive Surgery, Department of Surgery, Duke University Medical Center;* [2]*Department of Biomedical Engineering, Duke University, Pratt School of Engineering, Durham, North Carolina;* [3]*Department of Plastic Surgery, The University of Texas MD Anderson Cancer Center, Houston, Texas, USA*

Correspondence: Dr Howard Levinson, DUMC 3181, Division of Plastic and Reconstructive Surgery, Department of Surgery, Duke University Medical Center, Durham, North Carolina 27710, USA. Telephone 919-684-8661, fax 919-681-7340, e-mail howard.levinson@duke.edu

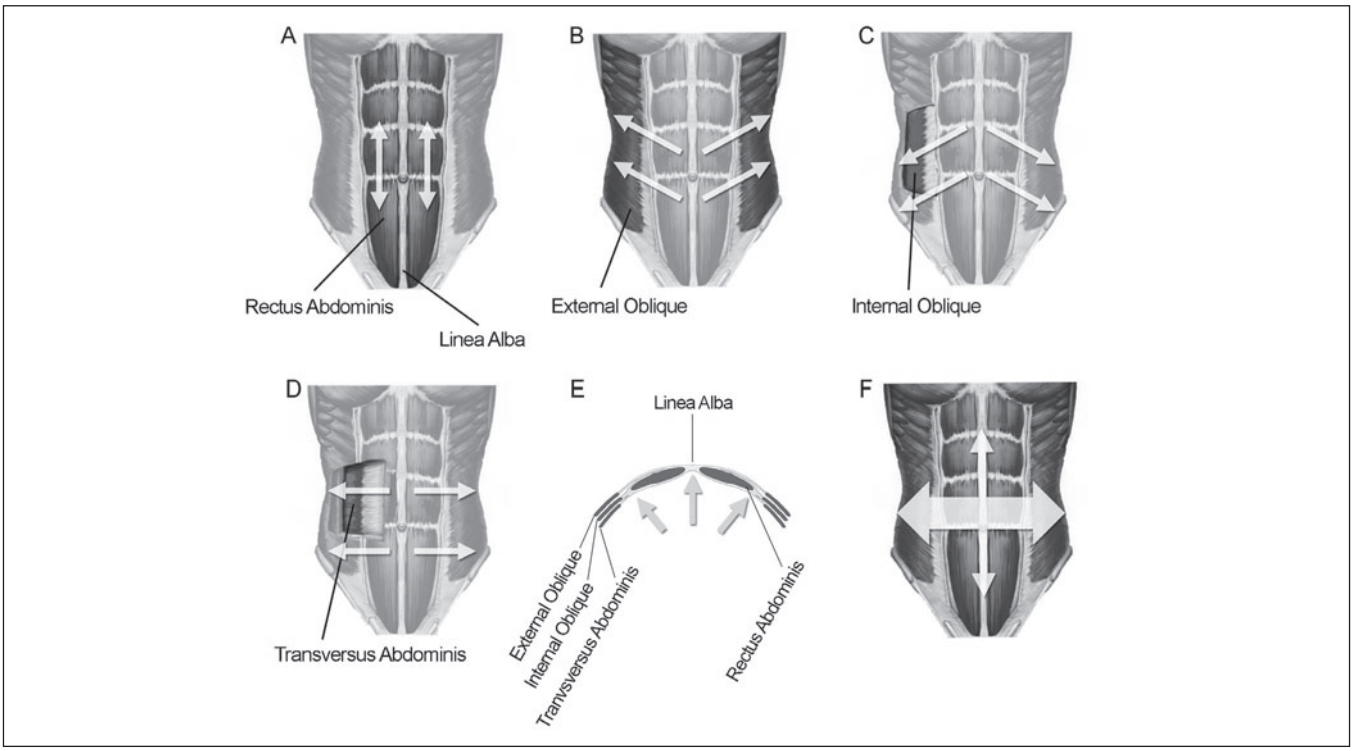

Figure 1) *Vectors of force in the abdominal wall. Contraction of the rectus muscles (***A***) does not promote hernia formation. Contraction of the external oblique muscles (***B***), internal oblique muscles (***C***), and transversus abdominis muscles (***D***) pull the rectus muscles apart and promote hernia formation. A cross sectional view (***E***) demonstrates the forces acting on the linea alba in response to increased intra-abdominal pressure (eg, coughing). Forces in the transverse direction are reportedly twice as much as the forces in the longitudinal direction (***F***) (134)*

together in the midline, a components separation may be helpful. Components separation is the partial release of the abdominal wall fascia that connects the oblique muscles with the rectus muscles (18). In patients in whom the rectus muscles still cannot be brought to the midline, a bridged mesh repair is required. Bridged mesh repairs have demonstrated higher recurrence and complication rates compared with nonbridged repairs and, are therefore, suboptimal, particularly with biologic mesh (19,20). Outcomes are significantly improved with a mesh reinforced repair, in which the fascial edges are closed completely over the mesh.

The abdominal wall is exposed to multiple forces that contribute to hernia formation (Figure 1). These forces result from contraction of the internal oblique, external oblique and transverse abdominis muscle groups, as well as increased intra-abdominal pressure. The rectus muscles are the only muscle group of the anterior abdominal wall that contracts in a cephalad-caudal direction, which probably does not contribute to hernia recurrence.

CLASSES OF MESH

For the purpose of simplification and uniformity in the present review, all materials used to support hernia repairs are referred to as 'mesh'. Meshes can be divided into two broad classes: synthetic and biologic. Synthetic meshes are either nondegradable or degradable, while biologic meshes are all degradable. For the purposes of the present review, the term 'degradable' is used for meshes that, at least in part, dissolve or remodel over time and are replaced by either scar tissue or regenerative matrix. The different classes of surgical mesh along with their relative advantages and disadvantages are listed in Table 1.

The synthetic nondegradable meshes, sometimes referred to as 'classical' or 'traditional' meshes, are generally the least expensive. The earlier materials used for these meshes – perlon and nylon – were later abandoned because perlon caused intense inflammatory responses and nylon was shown to degrade in the long-term (21). Currently, nearly all synthetic nondegradable meshes are made from one of three basic materials: polypropylene, polyethylene terephthalate polyester or expanded polytetrafluoroethylene (ePTFE) (22). The characteristics of the different types of synthetic nondegradable mesh are presented in Table 2.

Synthetic degradable materials were intended to reduce adhesions and provide a safe alternative for placement in infected fields (Table 3). Vicryl (Ethicon, USA) and Dexon (American Cyanamid Co, USA), for example, are used in open abdominal wounds. The drawback to these meshes, however, is that they degrade within one to three months and are associated with high recurrence rates (23-27). To overcome early degradation, newer synthetic biomaterial meshes have been developed. For example, Gore Bio-A (WL Gore and Associates, USA) mesh degrades in six months and has been shown to reduce recurrence rates, infection and pain (23,28,29). Phasix (Bard Davol Inc, USA) (23,30) and Tigr Matrix (Novus Scientific, USA) (31-33) also degrade over several months and are useful in hernia repair, as has been demonstrated in preclinical animal (23,31,32) and human pilot (33) studies. The long-term effectiveness of these newer synthetic degradable meshes remains to be tested in clinical practice.

Biological meshes were used for hernia repair because they were believed to promote regeneration, rather than scarring, and because they could also be used in contaminated or infected fields (34). Biological meshes are typically manufactured from decellularized human, porcine or bovine dermis; bovine or equine pericardium; or porcine intestinal submucosa (Table 4) (35). The most commonly used biological meshes include Alloderm (LifeCell, USA) (allogenic dermis collagen), Permacol (Medtronic, USA) (cross-linked porcine dermis collagen), Strattice (LifeCell, USA) (non-cross-linked porcine dermis collagen), and Surgisis (Cook Biodesign, USA) (porcine intestine collagen). Alloderm is more expensive (36) and, in general, human dermal meshes have a higher recurrence rate than xenogenics (37). The porcine dermis collagens have a slightly better side effect profile than Alloderm and Surgisis, demonstrated by lower rates of seroma formation, lower total surgical morbidity (38), lower failure rates, and longer

TABLE 1
Classes of mesh with their relative advantages and disadvantages

Class of mesh	Advantage(s)	Disadvantage(s)
Synthetic		
Non-degradable	Inexpensive	Not recommended for infected fields
	Low recurrence rates	Higher rates of infection, discomfort, and adhesions
Degradable	Better side-effect profile than non-degradables	High recurrence rates for older meshes
	Lower cost than biologicals	Insufficient evidence for newer meshes
Biological		
Degradable	Can be used in complex/ infected fields	High recurrence rates
		Expensive

TABLE 2
The materials used in synthetic nondegradable mesh

Material	Mesh	Characteristics
Polypropylene	Prolene Marlex Parietene* Surgipro*	Rigid, inert, used in most woven prostheses
Polyethylene terephthalate polyester	Dacron Mersilene†	Elastic, hydrophilic, also available as large-pore woven mesh
Expanded polytetrafluoroethylene (ePTFE)	Gore-Tex‡ Teflon	Rigid, hydrophobic, low integration decreases risk of adhesions

**Covidien-Medtronic USA; †Ethicon, USA; ‡WL Gore and Associates Inc, USA*

time to failure in contaminated or infected fields (39). Porcine materials are easier to manufacture than allomatrices: they can be harvested in larger and more consistent sheets, and harvesting conditions can be better controlled. Porcine acellular dermal matrices do have drawbacks, however, such as requiring modifications to curb the intense immune response (40). Modifications can be achieved through chemical cross-linking of collagen fibres, as well as enzymatic removal of antigenic groups in the collagen (which enables the use of non-cross-linked porcine materials)(40). Interestingly, cross-linked porcine dermis meshes are associated with a heightened foreign body reaction and pronounced early inflammatory response (41,42), while non-cross-linked porcine meshes demonstrate fewer adhesions and complications (40). Although biological meshes are routinely used in infected fields, their high costs remain a barrier to widespread use (43). In addition, there is insufficient evidence in the literature regarding the advantages of biologic meshes over synthetic meshes in hernia repair (44-46).

Composite meshes consist of two or more distinct components and were developed to improve the side effect profiles of meshes. Many composite meshes are 'biface implants' – meshes with a porous external surface to encourage tissue integration and a smooth microporous internal surface to prevent adhesions when placed in contact with viscera. The external surface generally consists of a nondegradable synthetic material, while the visceral surface can be any combination of degradable or nondegradable, synthetic or biological materials, such as polyglactin, collagen, polyglecaprone, cellulose, titanium, omega-3, monocryl, polyvinylidene fluoride and hyaluronate (47,48). Another group of composite meshes are not biface, but rather consist of a nondegradable synthetic mesh with a temporary barrier coating (48). Temporary barrier coated meshes have a barrier coating that is degradable and consists of a material that discourages adhesion formation, usually hydrophilic coatings such as collagen.

TABLE 3
Synthetic degradable meshes(23)

Material	Mesh	Degradation time
Polyglactin	Vicryl*	1–3 months
Polyglycolic acid	Dexon†	1–3 months
Polyglycolic acid/trimethylene carbonate	Gore Bio-A‡	6 months
Poly-4-hydroxybutyrate	Phasix§	12–18 months
Polyglycolide/polylactide/ trimethylene carbonate	Tigr Matrix¶	Includes two different fibre compositions; partially degrades in 4 months, completely degrades after 3 years

**Ethicon Inc, USA; †American Cyanamid Co, USA; ‡WL Gore and Associates Inc, USA; §Bard Davol Inc, USA; ¶Novus Scientific, USA*

TABLE 4
Biological mesh materials

Mesh	Examples
Allogenic	
Human dermis	Alloderm (LifeCell, USA)
	Allomax (Bard Davol Inc, USA)
	FlexHD (Ethicon, USA)
Xenogenic	
Porcine dermis	Permacol (Medtronic, USA)
	Collamend (Bard Davol Inc, USA)
	Strattice (LifeCell, USA)
	XenMatriX (Bard Davol Inc, USA)
Porcine intestine	Surgisis (Cook Biodesign, USA)
	Fortagen (Organogenesis Inc, USA)
Bovine dermis	SurgiMend (TEI Biosciences, USA)
Bovine pericardium	Veritas (Synovis Surgical Innovations, USA)
	Tutopas (Mentor Corp, USA)
	Periguard (Synovis Surgical Innovations, USA)

Thus, they theoretically promote integration and prevent adhesion formation during the initial period of implantation and then become a regular synthetic nondegradable mesh after the coating degrades. Examples of composite mesh currently on the market include Vypro (Ethicon, USA) , Parietex composite (Medtronic, USA), Composix (Bard Davol Inc, USA), Proceed (Ethicon, USA), Dynamesh (FEG Textiltechnik, Germany), Sepramesh (Bard Davol Inc, USA), Ventralight ST (Bard Davol Inc, USA), Ultrapro (Ethicon, USA), Ti-mesh (Medtronic, USA) and C-Qur (Atrium Medical, USA).

TISSUE ENGINEERING PRINCIPLES OF MESH DESIGN

The principles of functional tissue engineering (49) were originally developed to serve as a guide for designing implants that replace or repair body structures with important biomechanical functions. These principles include measuring the mechanical properties of normal tissue, prioritizing and selecting the most important physical properties of the tissue as they relate to the pathophysiology of disease, and engineering materials to overcome the current hurdles and complications. The following discussion presents some of the most important properties considered in hernia mesh design and manufacturing (Table 5).

One useful concept to consider through the following discussion is the difference between 'knit' and 'woven'. With knitting, a continuous filament is looped around another; while in weaving, a series of parallel strands are alternately passed over and under another set of parallel strands (Figure 2). Knit fabrics are more porous and flexible, while woven fabrics usually exhibit the same mechanical properties in each axis. Synthetic meshes (with the exception of the foils, such as ePTFE) are generally knitted, not woven (50).

TABLE 5
Important properties of mesh

Property	Definition	Goals/recommendations (reference[s])
Biocompatibility	Capacity to be implanted without producing an adverse effect	Non-toxic material with lowest amount of immune reaction (all materials produce some degree of reaction)
Mechanical properties		
Tensile strength	Maximum stress that a material can withstand while being stretched before failing or breaking	At least 32N/cm in the strongest direction, at least 16N/cm in the weakest (59)
Stiffness (Figure 3A)	The extent to which a material resists deformation in response to force	Goals stated as measures of elasticity (currently no standardized range of values).
Elasticity (Figure 3B)	The tendency to return to original shape after being deformed; measured by the elastic modulus, the tendency to be non-permanently deformed in response to a force	At most 30% at 32N/cm (47)
Compliance (Figure 3C)	The amount of displacement or deformation in response to a unit force	Goals stated as measures of elasticity (currently no standardized range of values)
Porosity and weight		
Porosity	The percentage of mesh not occupied by mesh material	(Currently no standardized range of values).
Pore size (Figure 3D)	The area between mesh filaments	Pores >75 µm allow macrophage infiltration, neovascularization and incorporation (74); pores >1 mm prevent granuloma bridging for polypropylene mesh (75,76)
Effective pores (Figure 3E)	The circular area between mesh filaments not occupied by granulomatous tissue	Circular interfilament distance of 1 mm for polypropylene mesh (70)
Weight	Measure of mass per unit of area	(Currently no standardized range of values)
Degradation	Disappearance of the mesh material	6 months for scar tissue to reach its maximum strength; (23,88,135) for adhesion formation the timeframe is unclear (128)
Constitution	The structural form of the mesh, including monofilament, multifilament, or foil structures	Monofilament mesh is preferable to multifilament mesh, due to a better side effect profile regarding foreign body reaction and infection
Anisotropy (Figure 3F)	The degree to which mechanical properties differ in response to applied loads in various directions; measured by the ratio between the elastic moduli in each axis for a given mesh	If mesh is anisotropic, its directionality must be acknowledged to address the forces it is subject to (currently no standardized range of values)

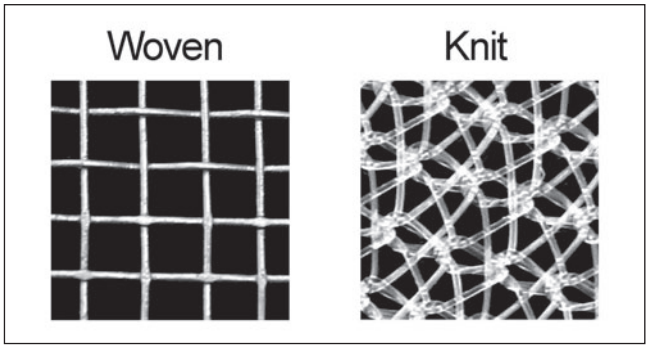

Figure 2) *Differences between woven and knitted fabrics. Woven fabrics consist of a series of parallel strands alternately passed over and under another set of parallel strands. Knit fabrics, such as the polypropylene mesh shown, consists of continuous filaments that are looped around one another*

Biocompatibility

The biocompatibility of mesh is dependent on a multitude of variables and is quantified in terms of the degree by which the material induces a foreign body reaction. Quantification includes measurement of the number of inflammatory cells (macrophages and granulocytes) present in the vicinity, granuloma size, vascularization, collagen deposition and mesh migration (51). Essentially all materials used in mesh development are chemically and physically inert, nonimmunogenic and non-toxic, yet none are biologically inert and all, including the biological meshes (52), trigger an array of adverse events, including a foreign body reaction (53). The predominant hypothesis for the foreign body reaction in inert nonimmunogenic materials is the protein absorption theory, in which proteins nonspecifically adhere to the material surface and subsequently lose patterns in their tertiary structure, revealing hidden binding domains that elicit an immune response (54). The proteins that adhere to the foreign body depend on the material and frequently include immunoglobulins, C3, fibrinogen and factor XII. It has been proposed that the difference in adsorption determines the differences in foreign body reactions. Subsequently, immune cells are recruited and giant cells form and establish granulomas around the foreign material. Ultimately, a fibrotic capsule forms around the foreign material (55,56). Of the materials commonly used as mesh, polypropylene may elicit the strongest foreign-body reaction (56). Additionally, multifilamentous polypropylene mesh may promote added fibrosis compared with monofilamentous polypropylene (57).

Mechanical properties

Tensile strength is probably the most commonly discussed mechanical property of mesh. Tensile strength is defined as the maximum force per cross sectional area that the material can withstand before failure or break (47). Force per cross sectional area is known as 'stress' and is measured in units of pressure, Pa or N/cm^2 (58). Because meshes are produced with a standard thickness, sometimes tensile strength is presented as N/cm width of mesh, omitting the value of the thickness, which is presumed a fixed amount. Obviously, the ultimate tensile strength needs to be adequate to withstand the amount of force that is exerted on the abdominal wall. Most commercially available meshes exceed the required tensile strength to withstand the physiological forces of the abdominal wall (59). Nonetheless, mechanical failure of synthetic permanent mesh has been reported in the literature and appears to be exclusive to lightweight meshes (60-63).

Elasticity, compliance and stiffness are terms that are frequently confused or inaccurately used interchangeably. Elasticity is defined as the tendency of a material to return to its original shape after being deformed and is measured by the elastic modulus. The elastic modulus is derived from the slope of the stress-strain curve and, depending on its application, can be measured on the initial part of the curve or the part that has the greatest functional importance. Elasticity is also expressed

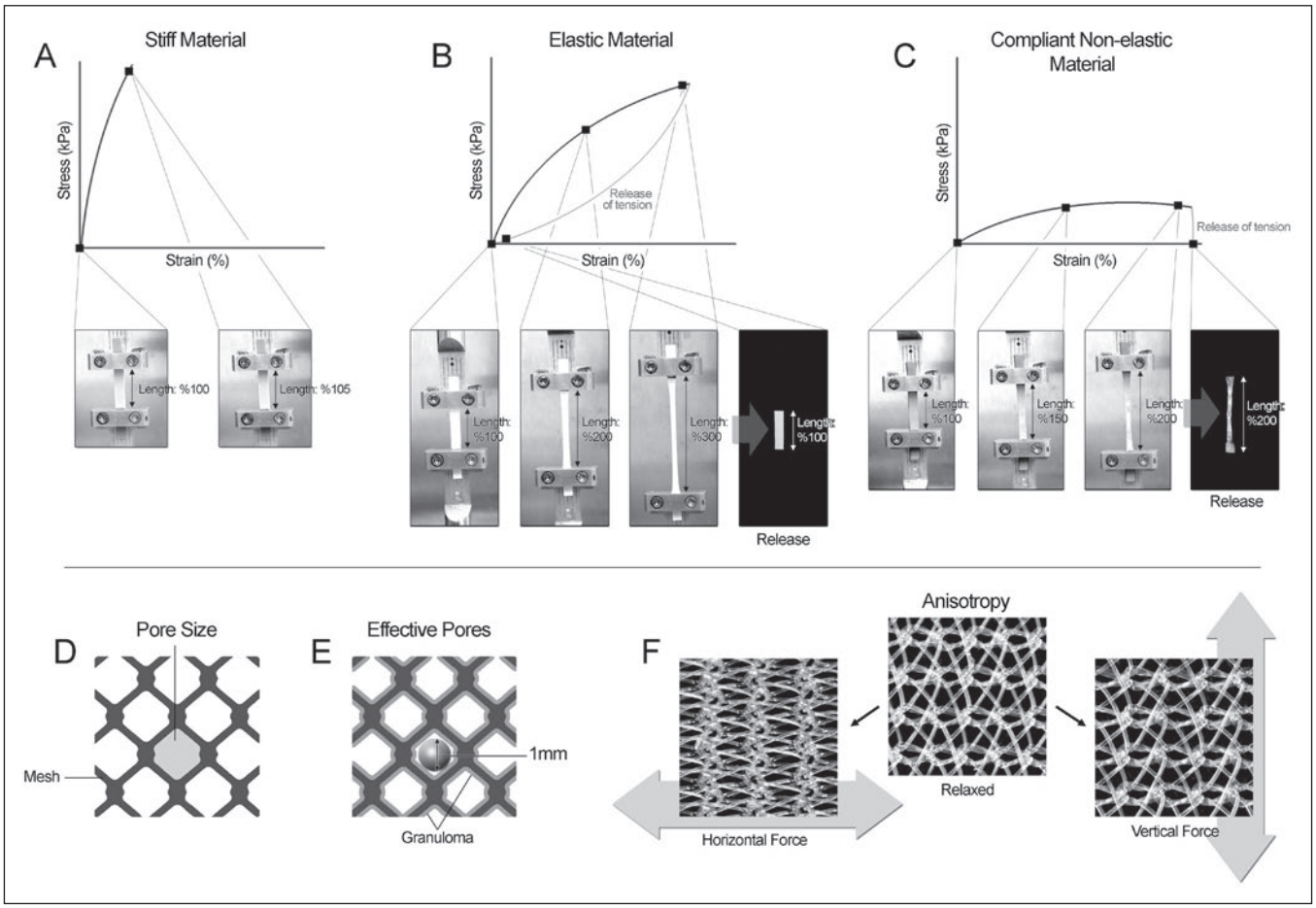

Figure 3) A *to* **C** *The difference between stiffness, elasticity and compliance. A stiff object (***A***) does not easily undergo deformation by force. An elastic object (***B***) will return to its original form when tension is released, up to the point where it undergoes plastic deformation. This point is considered to be the ultimate tensile strength of the object as opposed to the point of complete tearing. A compliant object that is not elastic (***C***) will deform readily and will not return to its original length.* **D** *and* **E** *Pores and effective pores. Pore size refers to the area between mesh fibres. Effective pores refer to pores that do not become occupied with granuloma tissue. This is frequently measured by pores that can fit spheres of a specific diameter (eg, 1 mm for polypropylene).* **F** *Anisotropy. This figure shows a polypropylene mesh when subject to force pulling in two perpendicular directions. When pulled in one direction, the mesh demonstrates minimal displacement, but when subject to force in the other axis, the displacement is evident*

as the amount of displacement in response to a specific measure of stress. It is an important property because meshes that are stretched out but do not return to their original size, will likely lead to recurrent hernias. The natural elasticity of the abdominal wall has been estimated to be approximately 38% at 32N/cm, and an elasticity >30% at 32N/cm may allow for more stretching than the normal abdominal wall would permit and, therefore, may not be suitable for a functional repair (47). On the other hand, a mesh with low elasticity would restrict abdominal wall distention, resulting in pain and mesh failure. It has been suggested that the lowest range for mesh elasticity is between 4% and 15% at 16N/cm (64).

Stiffness is defined as the extent to which an object resists deformation in response to an applied force and is the inverse of compliance. Overly stiff materials are more likely to dehisce from the abdominal wall and cause pain when the patient moves. Some have described mesh stiffness as the quotient of the maximum load and strain at the maximum load, but stiffness and compliance are not common measurements (59).

Pore size and weight

Pore size and weight are key aspects of mesh design, particularly with the more recently developed large-pore lightweight mesh (65-68). Pores <10 mm generally impede human cellular penetration and tissue ingrowth (69). Pore sizes ≤75 mm may hinder the access of antimicrobial agents and host immune cells to bacteria, thus, predisposing the material to bacterial colonization and infection. Such meshes are sometimes referred to as microporous meshes, as opposed to macroporous meshes with pore sizes >75 mm (70). The ePTFE foils (eg, Dualmesh [WL Gore and Associates, USA]) are the only microporous synthetic meshes and as such, frequently require removal when infected (71-73). As the pore size increases to 100 µm to 300 µm, neovascularization and tissue integration are frequently observed, but granuloma bridging becomes a concern (56). Granuloma bridging, or the coalescence of the foreign body response around mesh fibres, can clog the pores and prevent further tissue integration (Figure 4) (56). In polypropylene meshes, when pore sizes are <1 mm, granulomas can become confluent, encapsulate the mesh and create a stiff plate with reduced flexibility (70,74-76).

Although it was previously believed that large pore size would delay incorporation (77), this has not been observed in practice. In fact, the opposite has been described, in which large-pore meshes (with lower surface-area-to-volume ratios) result in a milder foreign-body reaction. The trade-off, however, is that reduced mesh material results in a base mesh with reduced strength.

Weight is another factor in mesh design. Mesh weight is partially dependent on polymer weight (74) but is mainly a function of pore size (75). With greater pore sizes, less material is used to construct the mesh, and mesh weight is reduced. In general, lightweight meshes tend

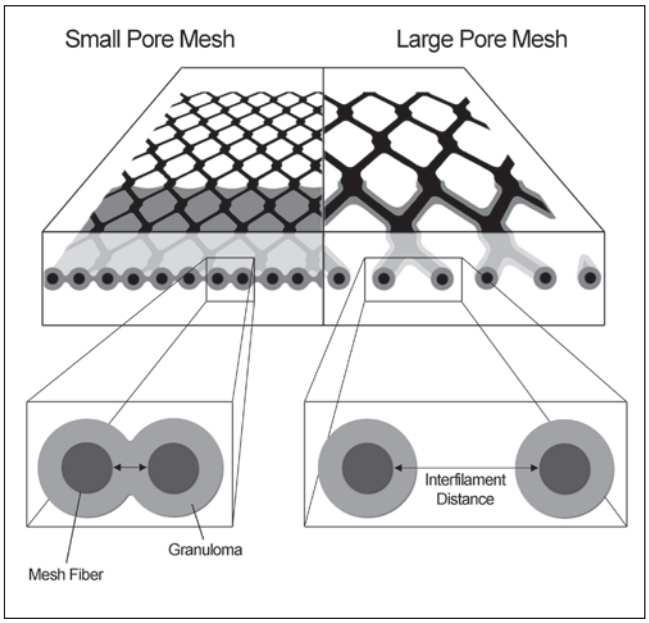

Figure 4) *Granuloma bridging. When pores are small, granulomas become confluent, leaving no remaining effective pores. In large-pore meshes, granulomas surround the mesh fibres but do not occupy the entire pore*

to weigh approximately 33 g/m^2, while heavyweight meshes tend to weigh approximately 100 g/m^2 (47,74). The rationale for regarding weight as an independent variable from pore size is the hypothesis that lighter weight meshes will have a smaller foreign body burden (78) and a smaller biomaterial surface area (79) and, thus, should elicit a less intense foreign body reaction. Some studies confirm this effect in practice and claim that lower weight results in fewer complications, while others do not (63,80). Specific recommendations regarding the ideal mesh weight remain to be determined. In addition, while some have attempted to classify mesh based on weight, such attempts and their cut-off points have not been completely supported by the evidence (70).

In practice, large-pore lightweight meshes are reported to have a similar profile to small-pore heavyweight meshes (81,82). At least one study demonstrated higher rates of shrinkage for large-pore lightweight mesh compared with small-pore heavyweight meshes (83). Some studies have suggested that large-pore lightweight meshes result in superior tissue integration (84), better elasticity (85) and a lower incidence of pain (86), while other studies report a higher recurrence rate for large-pore lightweight mesh in laparoscopically repaired groin hernias, especially in larger hernias (87).

Degradation

Degradation, defined as the disappearance of mesh or gradual decline in its mass, can be desirable or undesirable. In meshes that are degradable, the goal is to have the mesh last until scar or regenerative tissue replaces it and matures to maximum strength. From early experiences with Vicryl and Dexon, it is known that a three-month time frame for degradation would be inadequate (23-25). Recent data suggest a degradation time of six months could be successful, as evidenced by the studies that have demonstrated adequate outcomes with the Gore Bio-A mesh (23,28,29). This is analogous to studies in skin wound healing, which suggest that wounds regain 80% of their original strength by six months (88). However, the long-term recurrence rate of the Gore Bio-A mesh remains high, ranging from 13% to 37.5%, and it has been suggested that 12 months may be a better time frame for mesh degradation to ensure maturation of the scar tissue (28,29,89,90). This is where newer synthetic degradable meshes that have even slower degradation rates, such as Phasix and Tigr Matrix, could play a role.

In spanning defects that require the mesh to remain indefinitely to provide structural support, degradable mesh is contraindicated because the recurrence rate is nearly 100% (91). Unfortunately, even non-degradable mesh may slowly degrade. Polyester meshes are known to have the drawback of long-term degradation, which renders them unsuitable for long-term support (92). Recently, attention has even been drawn to the degradation of polypropylene, one of the most widely used materials in mesh development (93). It has been suggested that the degradation of polypropylene is accelerated with exposure of the material to heat during the manufacturing process (94). Early degradation of a mesh that is not intended to degrade may contribute to mechanical failure and hernia recurrence.

Another discussion regarding mesh degradation includes understanding what replaces the mesh once it has degraded: scar or regenerated tissue. For example, cross-linked porcine meshes are more antigenic and, are thus, replaced by scar, whereas non-cross-linked meshes are less antigenic and are replaced by regenerate tissue. Regenerate tissue exhibits a greater degree of cellular infiltration, degradation, deposition of extracellular matrix, neovascularization, lower inflammatory cell response, and less scar encapsulation, whereas scar tissue has limited host cell and vessel infiltration, more fibrotic matrix, and aligned collagen deposition (40,95).

Constitution

Synthetic mesh can be monofilament (mesh fibres are single filaments) or multifilament (mesh fibres consist of multiple filaments). Examples of multifilament meshes include Mersilene (a synthetic non-degradable multifilament mesh), Vicryl (a degradable multifilament mesh), and Vypro and Parietex (composite multifilament meshes) (74). Multifilament meshes are more pliable than monofilament meshes (96). Although some maintain that multifilament and monofilament mesh are comparable in terms of infection risks (97), the evidence suggests that multifilament meshes have higher infection rates and stronger foreign body reactions, due to the inaccessible crevices between the filaments, and larger surface areas (98-100).

Anisotropy

Anisotropy is the degree to which mechanical properties differ in response to applied loads in various directions and is quantified by the ratio between the elastic moduli in each axis for a given mesh (101). Almost all synthetic meshes exhibit various degrees of anisotropy. This is the result of synthetic mesh being a knit material as opposed to a woven fabric.

Because mechanical properties differ greatly based on directionality in knitted mesh, it has been recommended that anisotropy be identified and marked on the meshes to help surgeons orient meshes during implantation to optimize postsurgical outcomes (59,101,102). The rationale that the meshes should be aligned to maximally resist forces has yet to be tested or verified (101).

COMPLICATIONS

Hernia recurrence/infection

The most common complication following use of a surgical mesh is hernia recurrence (10,103-105). Fundamentally, recurrence is caused by early degradation of the mesh, early removal of the mesh (as necessary following infections) or mesh failure (Figure 5) (34,45,106). Mesh failure is caused by central mesh failure (mesh fracture) (60-63) or fixation/suture line failure (107). Central mesh failure almost always occurs in lightweight but not heavyweight meshes (60-63). Suture line failure is common and is typically reported as surgeon inexperience or fixation technique dependent. This is why so much effort is being made to find superior fixation techniques (108-111).

The rate of infection for open ventral incisional hernia repair is reported to be 6% to 10% (73). Patient- and procedure-related risk factors include obesity, chronic obstructive pulmonary disease, abdominal aortic aneurysm repair, previous surgical site infection, performance of other procedures via the same incision at the time of repair,

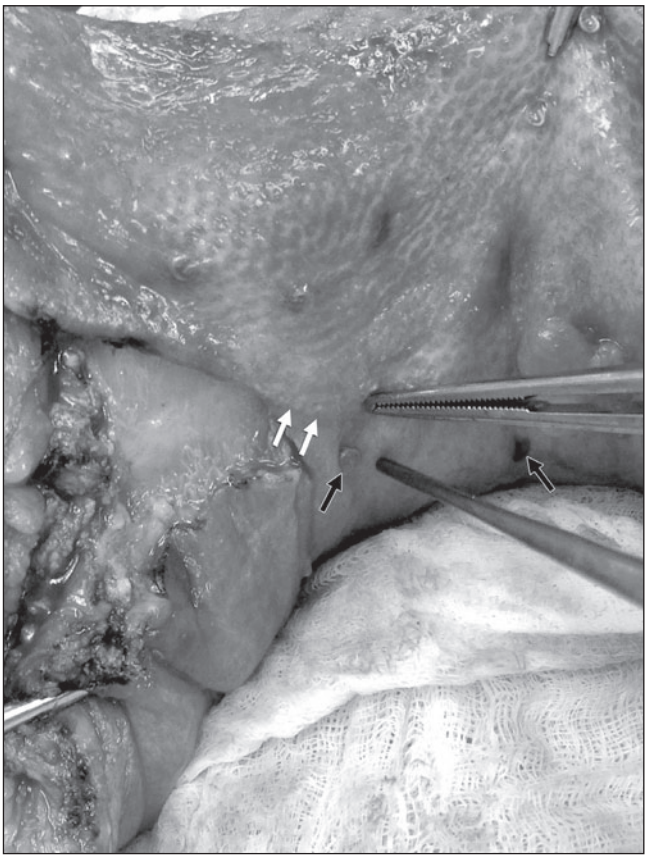

Figure 5) *Mesh failure in a patient with tack fixation. The mesh is seen up to the point shown by the white arrows. The black arrows show tacks without any surrounding mesh*

longer operative time, lack of tissue coverage of the mesh, enterotomy and enterocutaneous fistula (73). Mesh-related risk factors include the use of larger mesh sheets, microporous meshes or ePTFE mesh (73). Biological prostheses are commonly used in complex, contaminated, or potentially contaminated fields, but the exact reason why these biomaterials are safer to use is unclear (112). Controversy exists as to whether synthetic nondegradable meshes are also safe to use in an infected field (112-114). The concern is that once the infection is seeded on the nondegradable mesh, the infection will not resolve and an additional operation will be needed to remove the mesh. Some authors believe that there is a place for nondegradable meshes in infected fields, particularly because of the high costs of biologic meshes (115-117). Synthetic degradable meshes, however, have shown promise as a potential alternative to the biologicals for use in complex or infected fields (118).

One newly emerging concept is that of drug-eluting meshes, which allow for local delivery of antibiotics (119). Several studies have described methods wherein the prosthesis is coated with antibiotic containing solution; however, this may also alter its porosity, surface morphology and biomechanics (119-123). Antibiotic-eluting meshes could decrease bacterial contamination and biofilm formation. In addition, local drug delivery systems offer greater efficacy, prolonged drug activity, lower drug dose requirements, lower probabilities of antimicrobial resistance and generally lower toxicity (124,125).

Adhesion

For bridging meshes or when meshes are placed within the abdomen, viscera-mesh adhesion is a concern (Figure 6). Several studies have shown that biface (126) and barrier-coated (127) composite meshes are effective at reducing adhesion formation. A potential problem with

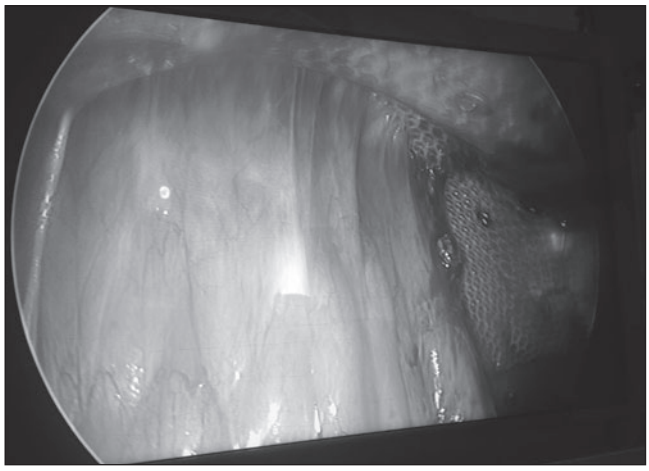

Figure 6) *Mesh adhesion*

temporary barrier coated meshes is that there is no specific timeline for adhesion formation (128); they can occur any time after mesh implantation. Stable hydrophilic coatings that do not degrade have been applied to address this issue, but this solution is still in its early stages and only limited animal model data exist (129). In general, ePTFE meshes have relatively low adhesion rates (130). Lightweight meshes have also been reported to exhibit low adhesion rates, which is presumably due to better integration and less foreign body reaction (131).

Postoperative pain

Postoperative pain is also a common complication of incisional hernia repair (132). While acute and early postoperative pain may be related to the type of mesh used, it is equally likely attributable to nerve damage from the operation (74). On the other hand, late-onset chronic postoperative pain is generally considered to be a complication of the mesh itself, and is most commonly associated with foreign body reaction and the resulting stiffness and shrinkage. In light of these data, some hypothesize that lightweight mesh or fully degradable mesh may decrease the risk for chronic pain (133).

CONCLUSION

The tissue engineering principle of 'replacing like with like' should be applied in abdominal wall reconstruction; however, abdominal wall properties are difficult to replicate due to its complex anatomy and dynamic requirements. In an effort to reduce ventral incisional hernia recurrence and the overwhelming associated costs, every effort should be made to choose the most appropriate mesh, as in certain settings, one type of mesh may be favoured over another. Manufacturers of mesh aim to improve their product by altering the properties described in the present article with each new product. Unfortunately, there is currently no ideal mesh, and surgeons must choose the 'best' available mesh given a clinical scenario. The present article presents the basic principles of mesh design to provide mesh users information on the many different types of meshes available, the properties of mesh and the critical issues facing the field of hernia repair.

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