World Journal of Gastrointestinal Surgery

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DOI: 10.4240/wjgs.v7.i10.226

World J Gastrointest Surg 2015 October 27; 7(10): 226-236
ISSN 1948-9366 (online)

EDITORIAL

Mesh implants: An overview of crucial mesh parameters

Lei-Ming Zhu, Philipp Schuster, Uwe Klinge

Lei-Ming Zhu, Department of Surgery, Tongren Hospital, School of Medicine, Shanghai Jiaotong University, Shanghai 200336, China

Philipp Schuster, FEG Textiltechnik mbH, 52074 Aachen, Germany

Uwe Klinge, Department for General, Visceral and Transplant Surgery, the University Hospital of the RWTH Aachen, 52074 Aachen, Germany

Author contributions: Zhu LM and Schuster P conceived the study and wrote the manuscript; Klinge U reviewed the manuscript and made valuable contributions to the intellectual content and structure of the paper.

Conflict-of-interest statement: Lei-Ming Zhu declares no conflict of interest. Philipp Schuster is an employee of FEG Textiltechnik mbH, Aachen, Germany. Uwe Klinge received fees for serving as a speaker, a consultant for FEG Textiltechnik mbH, and an expert in litigation. Uwe Klinge received research funding from the German government and commercial manufacturers.

Correspondence to: Dr. Uwe Klinge, Professor, Med, Department for General, Visceral and Transplant Surgery, the University Hospital of the RWTH Aachen, Pauwelsstraβe 30, 52074 Aachen, Germany. uklinge@ukaachen.de
Telephone: +49-241-8089352
Fax: +49-241-8082442

Received: April 10, 2015
Peer-review started: April 11, 2015
First decision: June 24, 2015
Revised: August 24, 2015
Accepted: September 25, 2015
Article in press: September 28, 2015
Published online: October 27, 2015

Abstract

Hernia repair is one of the most frequently performed surgical interventions that use mesh implants. This article evaluates crucial mesh parameters to facilitate selection of the most appropriate mesh implant, considering raw materials, mesh composition, structure parameters and mechanical parameters. A literature review was performed using the PubMed database. The most important mesh parameters in the selection of a mesh implant are the raw material, structural parameters and mechanical parameters, which should match the physiological conditions. The structural parameters, especially the porosity, are the most important predictors of the biocompatibility performance of synthetic meshes. Meshes with large pores exhibit less inflammatory infiltrate, connective tissue and scar bridging, which allows increased soft tissue ingrowth. The raw material and combination of raw materials of the used mesh, including potential coatings and textile design, strongly impact the inflammatory reaction to the mesh. Synthetic meshes made from innovative polymers combined with surface coating have been demonstrated to exhibit advantageous behavior in specialized fields. Monofilament, large-pore synthetic meshes exhibit advantages. The value of mesh classification based on mesh weight seems to be overestimated. Mechanical properties of meshes, such as anisotropy/isotropy, elasticity and tensile strength, are crucial parameters for predicting mesh performance after implantation.

Key words: Hernia repair; Hernia mesh; Incontinence mesh implant; Synthetic mesh; Mesh properties; Textile structure; Structure parameters; Mechanical parameters; Mesh weight; Synthetic raw materials

Core tip: Hernia repair is one of the most frequently performed surgical interventions that use mesh implants. This article evaluates crucial mesh parameters to facilitate selection of the most appropriate mesh implant based

on raw material, mesh composition, and structural and mechanical parameters. The structural parameters of the mesh, especially the porosity, are the most important predictors of the biocompatibility performance of synthetic meshes. Monofilament large-pore meshes exhibit less inflammatory infiltrate, connective tissue and scar bridging, which allows increased soft tissue ingrowth. The value of mesh classification based on the mesh weight seems to be overestimated. Other properties, such as the isotropy, elasticity and tensile strength, are crucial parameters for predicting the performance of meshes after implantation.

Zhu LM, Schuster P, Klinge U. Mesh implants: An overview of crucial mesh parameters. *World J Gastrointest Surg* 2015; 7(10): 226-236 Available from: URL: http://www.wjgnet.com/1948-9366/full/v7/i10/226.htm DOI: http://dx.doi.org/10.4240/wjgs.v7.i10.226

INTRODUCTION

Synthetic mesh implants are frequently used in many surgical interventions, especially in hernia repair. Mesh implants are composed of polypropylene (PP), polyethylene terephthalate (PET), expanded polytetrafluoroethylene (ePTFE), polyvinylidenefluoride (PVDF), and absorbable materials, such as polylactide (PLA), polyglycolic acid (PGA), polycaprolactone (PCL) and polydioxanone (PDO). Potential mesh-related complications include chronic infections, chronic pain and mesh rupture[1-3]. The reasons for chronic pain and the impact of mesh fixation in this context are controversial[4,5]. Chronic infections are favored by concomitant inflammatory and fibrotic reactions to the foreign body, hindering the local clearance from bacterial which leads to a chronic inflammatory wound with marked scarring, loss of compliance, mesh contraction, migration, physiochemical changes, seroma, infection, and in some cases, eventual mesh removal to resolve the problem[6]. A basic understanding of the physicochemical properties of meshes is essential for rational selection of the most appropriate device. This article evaluates the following crucial mesh parameters to facilitate selection of the most appropriate mesh implant: raw material, mesh composition, and structural and mechanical parameters (Figure 1).

The impact of mesh implants on clinical results is the current subject of much litigation in the field of stress urinary incontinence and pelvic prolapse, and some manufacturers were sued because of allegedly defective implants. However, many other factors besides mesh parameters must be considered in evaluations of the overall outcome of an intervention, including the patient's constitution, the selection of a proper operation technique and the operation performance, which are essential for the success or failure of a therapy.

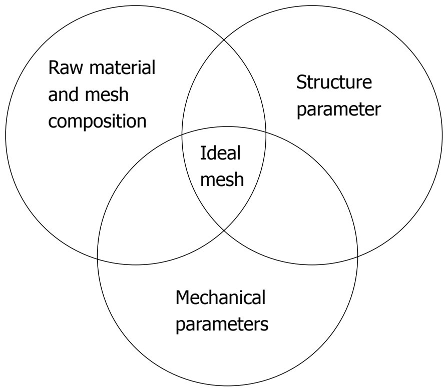

Figure 1 Crucial mesh parameters for selection of an ideal mesh.

BIOCHEMICAL FUNDAMENTALS

Implantation of a mesh triggers a foreign-body reaction, which plays a crucial role in the incorporation of the mesh into the host tissue. Incorporation of mesh into tissues is a complicated biochemical healing process. Implantation initiates an acute inflammatory cellular response that is initiated by protein absorption at the surface and attracts local inflammatory cells, such as macrophages, that converge to foreign body giant cells and eventually create a chronic wound around the mesh fibers. New blood vessels and collagen form around the mesh[7]. A relatively high level of macrophage invasion is detectable 20 min after mesh implantation, and these levels increase slightly and then decrease within 24 mo[8]. More than 80% of the cells in the mesh infiltrate positively express CD68, CD8, CD45R0 and vimentin, which indicates a mixture of cells of various origins and confirms the existence of multiple transition forms that are involved in the inflammatory response[9]. Complement and mast cell activation may also be involved in the mediation of local tissue responses to synthetic hernia meshes[10,11]. Cell migration is followed by collagen deposition, with an increase in the type Ⅰ to type Ⅲ collagen ratio over time[12]. The majority of tissue ingrowth and strength may be completed 2 wk after mesh implantation, but the final remodeling process is a very significant challenge[13]. Mesh-induced foreign body responses must be balanced to result in normal wound healing. Swift and adequate tissue ingrowth into the mesh results in superior biocompatibility and likely improved clinical performance. Intense or prolonged inflammation, bad infiltration, and immature collagen deposition result in scar plate formation, which can be accompanied by increased stiffness of the abdominal wall, shrinkage or deformation of the biomaterial, recurrence, adhesion, fistula or erosion of nearby tissue[14].

TEXTILE FUNDAMENTALS

Textile structures consist of mono- or multifilament fibers. Figure 2 shows the schematic appearance of

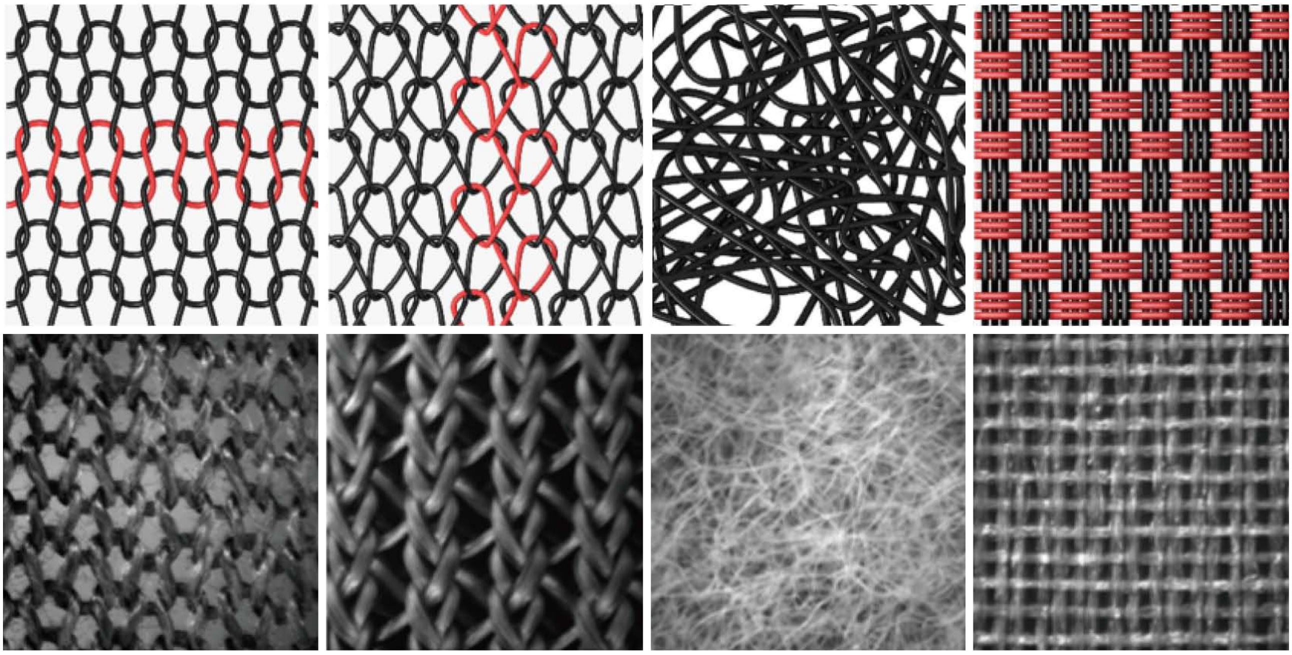

Figure 2 Textile structures from left to right: Knitted structure, warp-knitted structure, nonwoven structure, and woven structure.

Table 1 Definitions of the knit, warp-knit, nonwoven and woven textile structures

Textile structure	Definition
Knitted fabric	Knitted fabric consists of a number of consecutive rows of loops, called stitches. Knitted structures are manufactures from single yarn systems. Thus, knitted structures can be ribbed off. Trimming of knitted structures often leads to a complete falling apart
Warp-knitted fabric	Warp-knitted fabric consists of a number of consecutive courses of loops, called stitches. Warp-knitted structures are manufactures from multi yarn systems whereby the number of separate strands of yarn equals the number of stitches in a row. In contrast to knitted structures warp-knitted structures can be trimmed and sewed
Nonwoven fabric	Nonwoven fabric consists of non orientated or to a certain degree orientated staple or endless fibers. After the nonwoven formation the structure needs to be bonded which either is realised by mechanical, thermal or chemical bonding
Woven fabric	Woven structures consist of two distinct sets of yarns or threads which are interlaced at right angles to form a fabric

Table 2 Essential properties of the knit, warp-knit, nonwoven and woven textile structures

Textile structure	Porosity (macropores)	Elasticity	Mechanical behaviour	Trim-ability
Knitted fabric	++	++	Anisotropic	--
Warp-knitted fabric	++	++	Isotropic, anisotropic	++
Nonwoven fabric	-	-	Isotropic	++
Woven fabric	-	--	Isotropic	++

knitted, warp-knitted, nonwoven and woven structures. Table 1 provides definitions of these different textile structures.

Table 2 presents the general essential properties of these textile structures. These properties are adjustable in a wide range through the selection of production technology and through the specific settings of the production process parameters. Most textile mesh implants are warp-knitted because of the ability of these implants to provide large pores and elasticity under load. Warp-knitted meshes also do not lose material or structural strength at margins when trimmed to the size of the surgical need. Nonwoven meshes are used as mesh implants in exceptional cases.

RAW MATERIAL AND MESH COMPOSITION

Raw material

The polymer and fiber surface affect the inflammatory response within the granuloma. Most synthetic meshes use one of following raw materials: nonabsorbable materials, such as PP, PET, PVDF and ePTFE, or absorbable materials, such as PLA, PGA, PCL, PDO

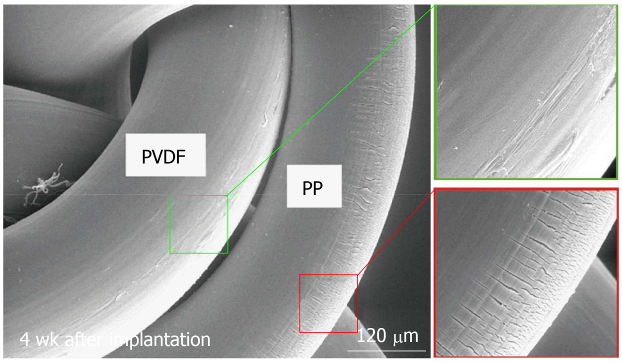

Figure 3 Comparison of the *in vivo* stability of the surface of polypropylene and polyvinylidenefluoride 4 wk after implantation[18]. PP: Polypropylene; PVDF: Polyvinylidenefluoride.

and PHB. These materials may also be used in combination with each other or a range of additional materials, such as titanium and hyaluronate. The foreign body reaction is fairly uniform regardless of the type of mesh implanted, but the different raw materials affect the extent of the reaction. PP meshes result in an intensified inflammatory reaction with deposition of more collagen fibers and significantly higher collagen type Ⅰ/Ⅲ ratios within the resulting scar neotissue compared with ePTFE meshes[15]. PET meshes induce the greatest foreign body reaction and longest-lasting chronic inflammatory response, which may be enhanced by the construction of PET fibers as a multifilament. Marked fibrosis and encapsulation surround ePTFE films[16]. PTFE is a more reactogenic material than PP, and it primarily stimulates the local production of pro-inflammatory cytokines. Therefore, the local anti-inflammatory effect of PP is less pronounced in comparison, but the inflammation persists for a longer time[17]. PVDF meshes produce a significantly reduced foreign body granuloma size compared with PP. PP is less stable than PVDF *in vivo*. Clear cracks in the surface of PP filaments have been detected 4 wk after implantation (Figure 3)[18]. These findings suggest that the raw material strongly influences the inflammatory and fibrotic responses.

Mesh composition

The primary aspects of mesh compositions are the use of different raw materials with or without surface coating in various textile designs.

Coatings may influence the degree of the inflammatory response. Nonabsorbable and absorbable materials are used for coatings. Absorbable materials are preferred if the coating provides a drug-eluting function. However, the degradation products may also influence the inflammatory response. A comparison of PP meshes, PP + polyglactin (PP + PG) meshes and PP + titanium (PP + TI) meshes demonstrated a reduced inflammatory reaction in the PP mesh group and increased reaction in the PP + PG mesh group. The PP mesh induced large early elevations in vascular endothelial growth factor, cyclooxygenase-2 and collagen levels, whereas the PP + PG mesh caused only small elevations in the levels of these factors. PP + TI meshes induced inflammatory response levels in between those of the other 2 meshes[19]. Human fibroblasts colonized on the macroporous PP side of a composite mesh made of two PP layers, but no cell growth occurred on the film PP side[20]. The suppressive effect of the mesh on the transforming growth factor β1 was more pronounced for partially absorbable materials compared with pure PP meshes, which suggests that a change in raw material composition and type affects the early biological reaction of connective tissue cells to the mesh[21]. Woven and nonwoven meshes have received less attention. Raptis *et al*[22] demonstrated that woven PP meshes became fully peritonealized intraperitoneally but generated thicker and more plentiful adhesions than nonwoven PP. PP nonwoven prosthesis are comparable to conventional warp-knitted meshes[23].

The textile design markedly influences the inflammatory reaction to the mesh. Using the best polymer in a poor textile design may lead to pronounced inflammation and scar formation. In contrast, an adequate tissue reaction may be achieved with a suboptimal polymer if the essential parameters of the textile design (*e.g.*, the filament structure and pore size) are considered. The particular type of mesh used in hernia repair may affect the wound healing response and clinical outcome[24].

STRUCTURE PARAMETERS

Pore characteristics

The characteristics of the mesh used - primarily the pore characteristics especially the collapse of pores under strain, amount of mesh material, prosthesis weight, and mechanical properties - crucially influence the dynamic incorporation. In 1997, Amid[25] identified mesh porosity as the decisive factor for risk of infection. Amid defined pores larger than 75 μm as macropores before large-pore meshes (3-5 mm) were developed. Klinge *et al*[26] evaluated a remarkable number of explanted meshes and found that the mesh porosity was the most important determinant of the tissue reaction and risk of scar entrapment. The pore size must be much larger than 75 μm to preserve tissue integration without filling the pores with scar tissue. A pore size > 1 mm is required for PP, and the pore size should be > 3 mm in cases of mechanical strain. Meshes with large pores exhibit less inflammatory infiltrate, connective tissue, fistula formation, calcification, and bridging (*i.e.*, the pores are filled by scar tissue) than meshes with small pores[27,28]. Granulomas normally form around individual mesh fibers as part of the foreign body reaction, but the term "bridging" describes the process whereby individual granulomas become confluent with each

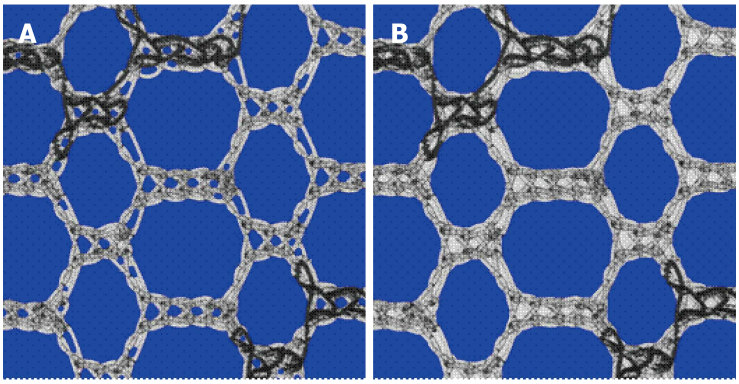

Figure 4 Comparison of the textile porosity (A) and the effective porosity (B).

other and encapsulate the entire mesh, which leads to a stiff scar plate and reduced flexibility[29]. A pore that is not completely filled by scar tissue is considered "effective" according to Mühl *et al*[30]. Therefore, large pore sizes preserve the "effective porosity" and thus avoid formation of scar bridges (Figure 4).

It is difficult to define a best pore size a priori because different raw materials result in different "effective porosities". Bridging of granuloma and encapsulation of the entire mesh is more likely for PP meshes with small pores (< 800 μm)[31]. In contrast, PVDF meshes do not exhibit bridging even for pore sizes of < 650 μm[32,33]. Klinge *et al*[26] characterized large-pore meshes using a textile porosity > 60% or an effective porosity > 0%. Pore shape may also determine integration. Lake *et al*[34] found that hexagonal pores resulted in the strongest tissue ingrowth, followed by square pores and diamond pores.

Mesh weight

Synthetic meshes may be classified as heavyweight or lightweight. The mesh weight depends on the polymer weight (raw material) and the amount of material used.

Coda *et al*[35] proposed a classification system based on the mesh weight that includes simple, composite and combined meshes. Meshes with weight per unit area of greater than 140 g/m² are defined as heavyweight meshes, meshes with weight per unit area in the range 35-70 g/m^2 are defined as lightweight meshes and meshes with weight per unit area in the range of 70-140 g/m^2 are defined as standard-weight meshes. Lightweight meshes generally contain less material and induce a less-pronounced foreign body reaction and decreased inflammatory response, which results in better tissue incorporation, increased prosthesis compliance, and decreased patient discomfort and pain. In an animal study, restriction of the abdominal wall mobility was significantly reduced and the inflammatory reaction and connective tissue formation were markedly diminished with lightweight meshes compared with heavyweight meshes[36]. Randomized prospective trials compared lightweight and heavyweight meshes for ventral hernia repair and found that they had equal outcomes in terms of ventral hernia recurrence[37]. Patients with lightweight mesh hernia repair exhibited better outcomes in terms of pain and seroma and an earlier return to activity[38].

Most current lightweight meshes have larger pores than heavyweight small-pore constructions[39]. However, mesh classification only in terms of weight disregards fiber and pore characteristics. Weyhe *et al*[40] considered the textile mesh construction, which they characterized in terms of the pore size and filament structure, as a more important determinant of foreign body reactions after implantation than absolute material reduction. This result attenuates the importance of mesh weight for the prediction of biocompatibility[41]. The advantages of lightweight meshes may be primarily related to their tendency to utilize a large pore size and/or monofilament.

However, an excessive reduction of mesh weight may also decrease the tensile strength. Lightweight meshes are sufficiently strong to resist abdominal wall pressure, but these meshes lose some burst strength compared with heavyweight meshes[32,42]. Experiments using small animals suggest that heavyweight small-pore meshes may withstand greater forces of scar contraction than large-pore lightweight meshes and may exhibit less shrinkage. Zogbi *et al*[43] found that lightweight PP mesh exhibited greater median shrinkage than heavyweight PP mesh in rats 7, 28 and 90 d after implantation.

MECHANICAL PARAMETERS

Mechanical properties are important parameters to consider when determining the suitability of a particular mesh for a specific clinical situation. However, surgeons typically implant meshes to provide maximum overlap over the defect with little regard for the mechanical properties of the mesh. Each synthetic mesh is composed of a unique combination of the material properties of the polymer and the textile design. The textile properties depend on the manufacturing process and the manufacturing process parameters.

The choice of raw materials determines a material's properties, which in many cases implies a combination of the properties of more than one raw material. These features ultimately determine the mechanical properties

Table 3 Definitions of mechanical mesh properties based on the definitions given by the American Society for Testing and Materials[44]

Property	Definition
Tensile strength	Tensile strength is the maximum force that can be applied to a mesh without tearing or breaking of the mesh. The tensile strength is measured in Newton (N) and is usually given in relation to the clamping width as Newton per centimeter (N/cm)
Burst strength	The burst strength is the maximum uniformly distributed pressure applied at right angle to its surface that a material will withstand under standardized conditions. The burst strength is given in pressure per unit area (Pa/cm²)
Elasticity (elastic elongation)	Elasticity (elastic elongation) is the property of a material whereby it changes its shape and size under the action of opposing forces (%), but recovers its original configuration when the forces are removed. In contrast, to the elastic elongation the plastic elongation indicates the elongation ratio which does not recover after unloading the structure
Stiffness	Stiffness can be expressed as ratio of steadily increasing or decreasing force acting on a deformable elastic material to the resulting displacement or deformation. Stiffness is a crucial aspect that reflects the drapablity of a textile structure, means the ability of a textile structure to be adapted to a 3-dimensional geometry

of the resulting mesh. An important consequence of the manufacturing process is the anisotropy of the tensile strength, elasticity, burst strength and stiffness. The American Society for Testing and Materials (ASTM) specification (D 4850 Terminology of textile structures) provides definitions of these properties (Table 3)[44].

The actual load on the abdominal wall is of major relevance for the selection of the suitability of meshes for use in ventral hernia repair. Different groups often perform simple tensile tests (N/cm), measurements of the inner abdominal pressure (Pa = N/mm²) or calculations of the abdominal wall tension (N/cm) using the Young-Laplace equation to characterize the native abdominal wall properties. The different measuring methods and different units should be considered when comparing these measurement results. Conversion of the inner abdominal wall pressure (Pa = N/mm²) (using the Young-Laplace equation) to the abdominal wall tension (N/cm) is only possible if the circumference of the patient is also provided. Use of the Young-Laplace equation requires a distinction between the sphere-like anatomy of the groin and the cylinder-like anatomy of the abdominal wall.

Hollinsky *et al*[45] measured the tensile load of the linea alba, the anterior and posterior rectus sheath, and scar tissue following median laparotomy in fresh cadavers and found that the tissue in the epigastric region ruptured at a mean horizontal load of 10 N/mm^2 in the linea alba and 6.9 N/mm^2 in scar tissue and at a mean vertical load of 4.5 N/mm^2 in the linea alba and 3.3 N/mm^2 in scar tissue. In earlier research, Williams *et al*[46] estimated the maximum force applied to the abdominal wall after hernia repair surgery as 22 N/cm in the cranial/caudal direction and 32 N/cm in the lateral direction. Cobb *et al*[47] investigated the intra-abdominal pressure using a transurethral bladder (Foley) catheter under different physical situations, including standing, sitting, bending at the waist, bending at the knees, performing abdominal crunches, jumping, climbing stairs, bench-pressing 25 pounds, arm curling 10 pounds, and performing a valsalva and coughing while sitting and standing, and identified a pressure of 22.7 kPa (171 mmHg) as the maximum pressure during coughing. Deeken *et al*[48] argued that stress in the transverse direction can reach levels of 47.8 N/cm in obese males with a large abdominal circumference. The true peak pressure in situations such as expectoration or sternutation in the abdominal wall was not fully addressed, but it is accepted that 22 N/cm in the cranial/caudal and 32 N/cm in the lateral direction are the maximum forces applied to the abdominal wall after hernia repair surgery[49]. A load of 16 N/cm is accepted as the maximum load in the groin because of the more sphere-like anatomy of the groin[50].

The natural elasticity of the abdominal wall at 32 N/cm is approximately 38%, with higher resilience in the horizontal direction than the longitudinal direction[45,46]. DuBay *et al*[51] indicate that the use of meshes in ventral hernia repair increases abdominal wall elasticity, which results in lower recurrence rates. Lightweight meshes exhibit an elasticity of approximately 20%-35% at 16 N/cm, but heavyweight meshes exhibit half of this elasticity (4%-15% at 16 N/cm), which may restrict abdominal distension[39]. An inappropriate mesh tensile strength, which results in an inappropriate ability of the mesh material to stretch, may potentially lead to poor functional results, with pain, hernia recurrence or prolapse. Elongation rates of greater than 30% indicate that these materials may stretch more than the native human abdominal wall. These meshes may not maintain functional repair, which could result in bulging or recurrence[48].

Tensile strengths of greater than 100 N/cm of conventional heavyweight meshes (*e.g.*, Prolene) are disproportionate and not necessary for effective repair[39]. Most synthetic meshes, even the lightest meshes, reach a tensile strength of at least 32 N/cm and are sufficiently strong. The mean burst strength and stiffness of lightweight meshes 5 mo after implantation in a pig was significantly less than those of heavyweight and middleweight meshes, but the burst strength for all meshes tested was much greater than the strengths measured for the abdominal wall fascia alone[32]. Bellón *et al*[52] demonstrated that the tensile strengths of lightweight and heavyweight meshes were comparable 90 d after implantation. However, Petro *et al*[53] recently reported 7 cases of mechanical failure or fracturing of lightweight monofilament polyester meshes after open incisional

Table 4 Essential properties of hernia meshes used for groin and abdominal wall hernia repair

Property	Recommendation
Tensile strength (abdominal wall)	22 N/cm (cranial/caudal) 32 N/cm (lateral)
Tensile strength (groin)	16 N/cm
Elongation	20%-40%
Orientation	No specific orientation for meshes with isotropic properties For meshes with anisotropic properties: orientation in the appropriate direction to match the physiological stretchability
Pore size	Depending on the used raw material and the foreign body reaction, respectively. To achieve a high effective porosity: for PP meshes a pore size ≥ 1000 μm should be used; for PVDF meshes a pore size ≥ 600 μm should be used

PP: Polypropylene; PVDF: Polyvinylidenefluoride.

hernia repair. Zuvela *et al*[3] and Lintin *et al*[54] reported central ruptures of low-weight PP meshes after initial sublay incisional hernia repair. These isolated case reports are insufficient to question the use of lightweight meshes in ventral hernia repair, but one should consider that the maximum initial tensile strength of synthetic meshes did not predict long-term strength after implantation[55]. Eliason *et al*[56] demonstrated that BardMesh, Dualmesh, and Prolene exhibited significantly reduced tensile strength, and BardMesh, Proceed, Prolene, ProLite, ProLite Ultra, and Ultrapro exhibited significantly increased permanent elongation after exposure to 1000 cycles of repetitive loading sequences that simulated changes in the intra-abdominal pressure. Mesh elongation also led to the loss of effective porosity in most meshes, which is an important aspect for scar formation and foreign body reaction[57]. Stiffness and breaking strength also vary widely among available meshes for hernia repair, and most meshes exhibit significant anisotropy in terms of their mechanical behavior. Pott *et al*[49] compared six meshes composed of different raw materials and different textile structures. All six mesh types exhibited differences in maximum tensile strength (11.1 ± 6.4 to 100.9 ± 9.4 N/cm), stiffness (0.3 ± 0.1 to 4.6 ± 0.5 N/mm), and elongation at break (150% ± 6% to 340% ± 20%) based on the load direction: the warp direction, or "longitudinal direction", *vs* the weft direction, or "orthogonal direction". Deeken *et al*[58] recently evaluated 13 mesh types that exhibited a wide range of mechanical properties. Some meshes were nearly isotropic, with nearly similar properties in the vertical and horizontal strain directions [C-QUR™, DUALMESH(®), PHYSIOMESH™, and PROCEED(®)], but other meshes were highly anisotropic (Ventralight™ ST, Bard™ Mesh, and Bard™ Soft Mesh). Some meshes exhibited a nearly linear behavior (Bard™ Mesh), but other meshes were non-linear, with a long toe region followed by a sharp rise in tension.

Meshes with different mechanical properties are treated as uniform and interchangeable, but it is important to understand the characteristics of the meshes to identify an appropriate mesh for each patient and place the mesh in an appropriate position to avoid mechanical mismatch, which may impair graft fixation, and enable optimized integration into the host tissue[59,60]. Therefore, surgeons may use meshes with isotropic properties regardless of the mesh orientation, but surgeons should pay attention to the orientation of meshes with anisotropic properties, which should be placed with their major elasticity in the appropriate direction to match the physiological stretch abilities (Table 4).

NEW DEVELOPMENTS

The evolution of meshes is not complete. New synthetic meshes are continuously developed, and new polymers and innovative coatings are continuously introduced. Ulrich *et al*[61] examined 3 new warp-knitted synthetic meshes composed of different polymers with different tensile properties, polyetheretherketone, polyamide (PA) and a composite, gelatin-coated PA (PA + G), in a rat model. All new materials exhibited better tissue integration, new collagen deposition and sustained neovascularization compared with PP meshes. Therefore, these new materials provide a promising alternative for future mesh developments. Meshes manufactured from native spider dragline revealed rapid cell migration, complete degradation, formation of a stable scar with constant tensile strength values and the highest relative elongation among standard biological and synthetic meshes[62].

Biosynthetic meshes are a possible cost-effective alternative to synthetic and biological meshes. Biodegradable polymers, instead of animal or cadaver tissue, provide a temporary scaffold for deposition of proteins and cells that are necessary for tissue ingrowths, neovascularization, and host integration[63]. Powell *et al*[64] reported good results in the early phase for "synthetic remodeling meshes" made from PGA/trimethylene carbonate in a study of 70 patients who underwent hiatal hernia repair. However, Symeonidis *et al*[65] used the same "synthetic remodeling mesh" in a pilot study of inguinal hernia repair and reported discouraging results, with a 38% recurrence rate after a mean follow-up of 2 years, which questions the general suitability of this mesh. Another fully absorbable mesh composed of knitted poly-4-hydroxybutyrate monofilament fibers, named the Phasix mesh, exhibited a strength that was 80%, 65%, 58%, 37% and 18% greater than the native abdominal wall at 8, 16, 32, and 48 wk post-implantation, respectively. The significant reduction of the polymers' molecular weight over time demonstrated successful transfer of load-bearing from the mesh to the repaired abdominal wall[66].

Configurations that include a metal component may also add new properties to standard synthetic meshes. Mesh shrinkage, migration, and configuration changes in the host tissue cause severe complications and discomfort after mesh implantation. There is no way to revise an implanted mesh postoperatively except for access to samples that have been explanted because of severe infection, chronic pain and recurrence. However, incorporation of small iron particles into the polymer provides an effective option for noninvasive revision using magnetic resonance imaging[67]. Another promising metal to improve mesh performance is nitinol. Nitinol-containing memory frame mesh is a valuable tool to achieve complete deployment in transinguinal preperitoneal repair for inguinal hernias that offers an acceptable morbidity and a low recurrence rate[68].

Coatings are another effective method to modify the properties of synthetic meshes. A titanium-coated PP mesh was associated with less postoperative pain in the short term, lower analgesic consumption and shorter convalescence compared with the Parietex composite mesh[69]. Intraperitoneal implantation of PP meshes is not recommended because of the likeliness of inducing intense adhesion and intestinal fistula. A PP mesh coated with poly(L-lactic acid) exhibited an additional property of anti-adhesion in a rat model[70]. Extracellular matrix-coated PP meshes attenuated the pro-inflammatory response with reduced cell accumulation, fewer foreign body giant cells and decreased collagen density without changes in the mechanical properties of the mesh[71,72]. Chitosan-coated PP meshes elicited preferential attachment of myoblasts over fibroblast attachment *in vitro*, which was associated with the restoration of functional skeletal muscle with histomorphological characteristics that resembled native muscle *in vivo*[73]. Degradable drug delivery coatings with incorporated antibiotics provide a specific approach to reduce post-surgical infections[74]. These promising laboratory and animal trial results may be incorporated in clinical practice in the future.

The use of electro-spun nanofibers of various polymers as tissue scaffolds in hernia repair has been an active research topic in recent years. Electro-spun materials feature three-dimensional nanofibrous structure with high surface-to-volume ratios and high porosity with high pore-interconnectivity that are similar to the native extracellular matrix. Drugs and growth factors for the prevention of incisional hernia formation have also been incorporated into electro-spun nanofibers[75]. Recent research revealed that PET and PET/chitosan electro-spun meshes performed well during incisional hernia surgery. However, the formation of foreign body granuloma in response to electro-spun structures was greater than when conventional meshes were used[76]. Further studies are required to elucidate the mechanisms that underlie the interactions between cells/tissues and nanofibrous materials.

CONCLUSION

Large-pore, monofilament, lightweight synthetic meshes are the current standard of practice. However, the risk of infection and other complications associated with the use of meshes are inevitable. An ideal synthetic mesh should consist of a monofilamentous large-pore structure with anisotropic mechanical properties that are similar to the native properties of the healthy host tissue and composed of a highly biocompatible raw material with long-term stability. An optimal mesh for intraperitoneal use must resist visceral adhesions to limit the risk of bowel obstruction and intestinal fistula. The use of innovative raw materials or coatings of currently available raw materials are promising approaches to realize these ideals. The individual response of the patient influences the local response after mesh implantation. Therefore, a thorough understanding of the biological processes of tissue formation and remodeling in the context of wound-healing processes after hernia repair is needed.

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P- Reviewer: Dmochowski R, Garg P, Vilallonga R **S- Editor**: Ji FF **L- Editor**: A **E- Editor**: Lu YJ

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