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Advanced Polymeric Surgical Sutures: From Material Design and Structural Engineering to Clinical Performance

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ABSTRACT

Polymeric sutures (absorbable or non-absorbable) are medical stitches designed to hold tissues together during the healing process after surgery or injury. They offer many advantages, such as biocompatibility, ease of handling and tying, good tensile strength, flexibility, and affordability. These sutures are categorized based on material and structural characteristics. Smart polymeric sutures can monitor and respond to biological conditions in real-time and self-tighten under body temperature. Various suture topologies exist, including monofilament, multifilament, braided, and barbed sutures. Different suture topologies offer varying mechanical and biological properties. Braided sutures have much more capillarity than non-braided sutures, facilitating fluid permeation through the suture thread, hence increasing susceptibility to tissue inflammation. Barbed structures provide tight closure of several skin layers without knot tying, while inhibiting bacterial adhesion and proliferation, reducing tissue irritation and the risk of wound infection. To transform sutures from a passive wound-closing device to an active therapeutic platform, sutures are loaded with drugs either by physical or chemical methods, depending on the drug's properties, suture material, and the desired release profile. Various natural and synthetic materials exhibit differing degradation rates, tensile strengths, and biocompatibility. Nowadays, synthetic polymers offer superior reliability and improved patient outcomes; thus, they replace natural materials. They are engineered to break down at a desired rate, which gives surgeons excellent control during healing. Also, the human immune system has a moderate response compared to natural materials (like catgut), resulting in less inflammation and discomfort for the patient. The future goal for surgical sutures is to transform them from passive closure devices into multifunctional, bioactive platforms that achieve many functions simultaneously, such as drug delivery, sensing, and enhanced healing, with improved mechanical properties, handling, and safety—all in a cost-effective and potentially sustainable manner. This means replacing the current "one-size-fits-all" approach with a drive for patient-specific solutions.

1. INTRODUCTION

Various methods exist for wound closure, including laser welding, skin adhesives, and staples [1, 2]. The predominant approach used is the utilization of sutures [3]. A suture is a thread of textile material, either synthetic or natural, used to ligate blood vessels and approximate tissues. It comprises a fiber with a metal needle affixed to one end of the fiber [4]. The primary objective of suture application in a clean incised wound is to align the wound edges until healing has advanced to withstand normal tensile pressures [3].

Throughout the millennia, several suture materials have been produced by scientists and used by doctors, dentists, and veterinarians. Original sutures were composed of organic materials, including silk and gut, which are produced by twisting strands of pure collagen. Among the vast array of materials now available, natural and synthetic polymers are the most commonly prioritized [5]. Contemporary sutures are predominantly synthetic, encompassing absorbable materials such as poly(glycolic acid) (PGA), poly(lactide-co-glycolic acid) (PLGA), poly(lactic acid) (PLA), polycaprolactone (PCL), poly(4-hydroxybutyrate) (P4HB), and poly-dioxanone (PDO or PDS), as well as non-absorbable polymers including poly(vinylidene fluoride) (PVDF), polyester (a copolymer of polybutylene terephthalate and polyglycol terephthalate), polypropylene (PP), poly(ethylene terephthalate) (PET), nylon, and polytetrafluoroethylene (PTFE) (Figure 1) [6].

Furthermore, stainless steel has been utilized as a suture material owing to its superior tensile strength and is employed in sternal closure, hernia repair, intestinal anastomosis, abdominal wound closure, and specific orthopedic treatments [7].

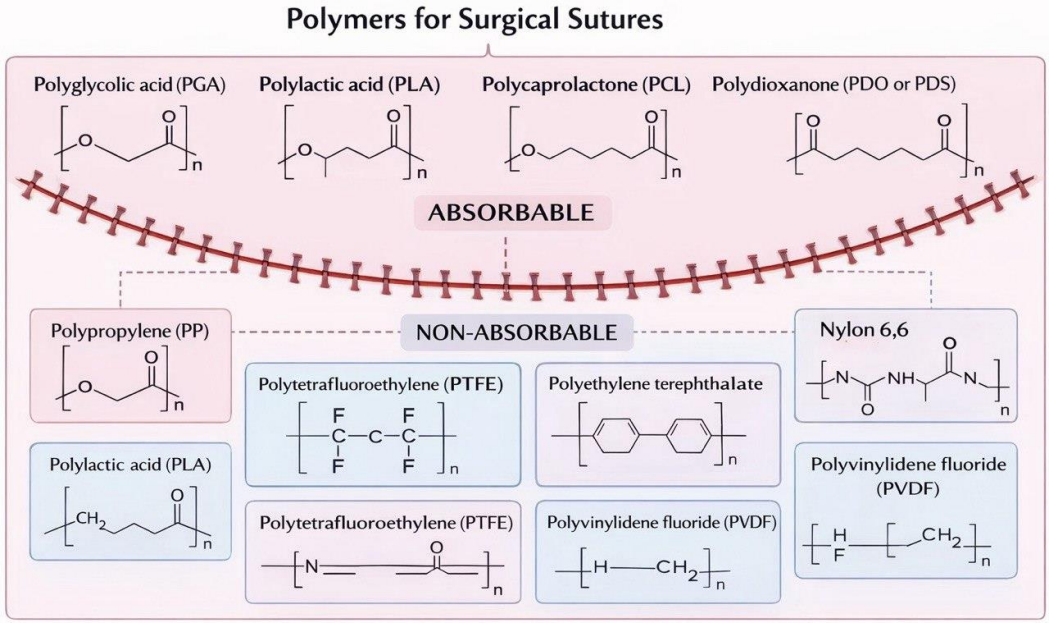

Figure 1. Some employed polymers for surgical sutures and their structures
Adapted from Afewerki et al. [6]

The fundamental property governing the performance of suture materials is their tensile strength, which can be tailored by modifying the fiber diameter and material composition [8]. In addition to mechanical integrity, several critical factors are considered in the selection of sutures, including high knot security, sterility, bioabsorbability, biocompatibility (ensuring minimal allergic or inflammatory reactions), and ease of handling and manipulation during surgical procedures [9]. Beyond the intrinsic features of the biomaterial, suture selection must account for clinical parameters, including the biological characteristics of the tissue, healing rates, the nature and severity of the wound, the patient's systemic health condition, the likelihood of postoperative complications, the surgeon's technical preference and expertise, and economic constraints [10, 11].

Additional performance considerations include the filament's capillarity, which affects fluid wicking and the potential for infection, tissue reactivity, and the rate of wound healing, all of which must align with the degradation profile of absorbable sutures to ensure optimal wound support and minimize complications. The overall aims of this study are:

(i) To provide a comprehensive overview of the fundamental properties, materials, and design principles of modern surgical polymeric sutures.

(ii) To critically analyze and compare the performance of different classes of absorbable and non-absorbable polymeric sutures.

(iii) To synthesize and present the latest advancements in "smart" and high-performance suture technologies.

(iv) To identify key challenges and limitations in current suture technology and clinical practice.

(v) To outline clear and promising future research directions for the next generation of surgical sutures.

2. HISTORY OF POLYMERIC SUTURES

The evolution and advancement of suture materials have spanned millennia, with their origins traceable to ancient civilizations. Historical records indicate that as early as 3000 B.C., ancient Egyptian practitioners utilized linen fibers to approximate wound edges. Subsequent developments included the use of catgut, introduced around A.D. 175 by Galen, who employed it in surgical procedures for Roman gladiators. Early suture materials were derived from natural sources, including flax, hemp, horsehair, human hair, pig bristles, and even anatomical components of insects like pincher ants. During the 19th and early 20th centuries, the widespread use of silk, cotton, and catgut dominated surgical practices. In 1869, Joseph Lister pioneered techniques to improve the sterility and performance of catgut sutures by impregnating them with chromic acid and introducing sterilization processes. Later, William Halsted emphasized the superiority of silk over catgut, leading to its predominance in surgical applications by the early 20th century [12, 13].

The mid-20th century marked the introduction of synthetic polymer-based sutures. During the 1940s, materials including nylon and Dacron, initially developed for other industrial purposes, were adapted for surgical sutures. The 1960s saw significant advancements with the work of Frazza and Schmitt, who developed synthetic absorbable sutures using polymers such as PGA, polyglactin 910, and PDO [12, 13].

Bayraktar and Hocienberge [14] conducted a comparative study on the knot performance of various suture materials, including silk, polyamide, polyester, and polypropylene. Their investigation demonstrated that knot integrity is significantly influenced by the suture's physical configuration—specifically, whether it is braided or monofilament. Braided sutures exhibit higher coefficients of friction due to inter-fiber mobility within the braid, thereby enhancing knot security and preventing slippage prior to material failure. In contrast, monofilament sutures possess smoother surfaces, resulting in lower knot-holding capacities and necessitating additional throws for reliable knot stability.

Lou et al. [15] developed absorbable surgical sutures composed of PLA. Their research confirmed PLA fibers' favorable biocompatibility, biodegradability, and mechanical strength. The sutures demonstrated optimal tensile strengths ranging from 3.1 N to 12.3 N.

This supports their potential for use in bone tissue

engineering and surgical applications requiring temporary wound support. Rethinam et al. [16] fabricated absorbable sutures from collagen derived from tannery solid waste. The collagen was blended with ethylene glycol, lyophilized to form a semi-solid paste, and processed into sutures. The resulting material exhibited high tensile strength (43.16 ± 1.03 MPa), excellent biocompatibility, and prolonged storage stability for up to six months. These features make the collagen-based sutures suitable for external and internal surgical interventions and wound healing. López-Saucedo et al. [17] utilized radiation grafting to functionalize polypropylene (PP) suture threads with N-vinyl imidazole (NVI), subsequently immobilizing silver nanoparticles on the surface. The antimicrobial activity of these modified sutures was validated against Escherichia coli and Staphylococcus aureus, with cytocompatibility correlating with the quantity of silver incorporated. Liu et al. [18] engineered a drug delivery system using PGA and PCL as carriers for ciprofloxacin, which were then coated onto PLA sutures. By adjusting the PCL/PGA ratio, they modulated the drug carrier's degradation rate and controlled the release kinetics of ciprofloxacin. Surface morphology analysis revealed that coating the PLA sutures with the PGA/PCL blend increased surface roughness and stitching resistance, offering the potential for targeted drug delivery in surgery applications.

Deng et al. [19] developed potassium-loaded sutures using a composite of PEG, PCL, chitosan, and keratin, fabricated through cost-effective hot melt extrusion. These drug-embedded fibers demonstrated homogeneous distribution, high thermal stability, and excellent mechanical properties, suggesting their applicability in soft tissue repair, including muscles, tendons, ligaments, and fascia.

Richard and Verma [20] introduced electrospun curcumin-loaded nanofiber yarns based on poly-L-lactic acid (PLLA) for future applications. These yarns provided a biomimetic three-dimensional architecture conducive to drug encapsulation, enhancing bioavailability and promoting tissue integration. Their sutures exhibited superior mechanical strength, sustained drug release, and improved antibacterial and anti-inflammatory properties, ultimately accelerating wound healing and tissue regeneration compared to conventional sutures.

Li et al. [21] reviewed various strategies for developing antimicrobial surgical sutures through functionalization techniques, including coating, melt spinning, wet spinning, grafting, electrospinning, and blending. Despite promising laboratory results, most antimicrobial sutures remain in preclinical or early clinical stages. The study emphasized the necessity for further innovation in this field to enhance infection prevention, support wound healing, and address manufacturing and material design requirements for clinical translation.

Alhulaybi [22] recently fabricated absorbable surgical sutures using PLA and PLA-chitosan composites via extrusion methods. These sutures demonstrated high elongation at break, consistent diameter control, and satisfactory knot security. Biocompatibility was confirmed using human skin simulators and in vivo rat models, with degradation studies showing 50% mass loss over 15 days, indicating controlled biodegradation appropriate for wound closure applications.

In conclusion, the continuous development of suture materials—from ancient natural fibers to modern polymer composites and drug-loaded systems—reflects the integration of materials science and biomedical engineering to address evolving surgical demands. Current trends emphasize biocompatibility, mechanical performance, drug delivery capabilities, and antimicrobial efficacy, promising significant advances in surgical wound management.

3. REQUIREMENTS OF SUTURES

Given the extensive range of available suture materials, a comprehensive understanding of their fundamental features is essential to ensure appropriate selection for optimal wound closure and surgical outcomes [12, 13]. Despite significant advancements in suture technology, no single material fulfills all the requirements to be deemed universally ideal. An optimal suture material would possess characteristics that allow it to be used in any surgical procedure, irrespective of tissue type or operative condition [1, 12]. Such a material would offer ease of handling, facilitate the formation of secure and reliable knots, exhibit superior tensile strength, and maintain mechanical integrity without eliciting adverse biological responses. Furthermore, it should neither impede the wound healing process nor act as a nidus for infection and should possess high visibility to aid surgical placement and removal [23].

The ideal suture must also demonstrate elasticity sufficient to accommodate tissue swelling (edema) and possess the ability to recoil in conjunction with wound contraction. Additionally, it should present an inhospitable surface for microbial colonization, be amenable to effective sterilization protocols, and display minimal tissue reactivity. To be clinically viable, such sutures should also be non-electrolytic, non-allergenic, non-carcinogenic, and economically feasible for widespread use [1, 12].

In clinical practice, permanent retention of sutures is rarely necessary or desirable. Prolonged presence of suture material within host tissue increases the risk of chronic inflammation and other undesirable tissue responses. Therefore, an ideal suture should maintain sufficient tensile strength throughout the critical phases of wound healing and subsequently undergo controlled degradation or absorption. The absorption process should occur promptly upon fulfilling its mechanical function without imposing excessive metabolic demands on the surrounding tissue [24].

4. CLASSIFICATION OF SUTURES

The selection of an appropriate polymer for surgical sutures necessitates careful consideration of both mechanical performance at the wound site and the material’s ability to minimize the risk of surgical site infections (SSIs). Optimal suture materials must exhibit the requisite tensile strength, flexibility, and handling features to support wound closure while offering biocompatibility to reduce inflammatory responses and the potential for postoperative complications. Suture materials are generally classified into two primary categories based on their origin: natural or synthetic.

Sutures are systematically classified according to several criteria:

(i) Absorption characteristics, distinguishing between absorbable and non-absorbable types;

(ii) Material origin, whether natural (for example, silk, catgut) or synthetic (for example, polyglycolic acid, polypropylene);

(iii) Yarn construction, encompassing monofilament, multifilament, twisted, or braided structures;

(iv) The presence or absence of dye, which can enhance visibility during implantation;

(v) Surface coatings, which may improve biocompatibility, reduce tissue drag, or confer antibacterial features;

(vi) Caliber (gauge or size): typically standardized according to established measurement systems [11].

4.1 Absorbable vs. non-absorbable sutures

Sutures are subdivided into absorbable and non-absorbable types, depending on their ability to undergo biodegradation and be resorbed by physiological processes. Natural, non-absorbable sutures, including those made from cotton, silk, and linen, have become less favored in contemporary surgical practice due to their propensity to act as substrates for bacterial colonization, thereby increasing the risk of SSIs [25]. On the other hand, natural absorbable sutures, primarily derived from collagen or catgut produced from animal intestines, have seen a decline in use owing to their inconsistent mechanical properties, including low tensile strength, and their tendency to elicit pronounced tissue inflammatory responses [26, 27]. In contrast, synthetic absorbable sutures offer controlled and predictable degradation kinetics, often through hydrolysis, and present a reduced risk of microbial contamination compared to natural alternatives [27]. These advantageous characteristics have positioned synthetic absorbable materials as the preferred choice for many surgical applications, particularly in procedures involving internal organs and deep tissues, where the integration of therapeutic agents, including antimicrobials or pharmaceuticals, into the suture structure further enhances their clinical efficacy [25].

Absorbable sutures preserve mechanical integrity during tissue restoration before biodegrading to eliminate the need for removal. In contrast, non-absorbable sutures sustain wounds permanently, especially in mechanically demanding situations. This is because non-absorbable sutures provide permanent support as they are made from chemically inert, synthetic materials (like PP or PS) that the body's fluids cannot break down. Instead of dissolving, the body simply walls them off with scar tissue, leaving the strong, intact suture in place to bear mechanical loads indefinitely. This makes them essential for structures under constant stress, such as heart valves or tendons, which require lifelong reinforcement.

In recent years, biodegradable polymeric materials have garnered significant attention in surgical applications due to their capacity for in vivo degradation, eliminating the need for secondary surgical procedures to remove the suture material. This characteristic minimizes the risk of postoperative infection, reduces patient morbidity, and enhances overall recovery outcomes. Absorbable sutures are typically characterized by their ability to retain no more than 50% of their original tensile strength within 60 days following implantation [28]. The mechanism governing the degradation and subsequent reduction in the mechanical performance of these sutures is primarily influenced by their composition; natural absorbable sutures generally undergo enzymatic degradation (proteolysis), while synthetic absorbable sutures predominantly degrade via hydrolytic cleavage of polymer chains [29]. This difference in degradation mechanisms stems directly from the fundamental chemical structure and origin of the materials. Natural absorbable sutures derived from biological tissues, primarily purified animal collagen (from intestines or tendons), have a chemical structure consisting of proteins, which are long chains of amino acids linked by peptide bonds. The body is already fully equipped to recognize and break down proteins. It has a dedicated system of enzymes called proteases, whose specific job is to cleave peptide bonds. For example, when catgut is implanted, the body identifies it as a "foreign protein" and launches a standard inflammatory response. Immune cells are recruited to the site and release these proteases, which systematically "digest" the suture material. In contrast, synthetic absorbable sutures are human-made polymers engineered in a lab. Their backbone consists of ester functional groups (-COO-) linking together monomer units. The body does not have specific enzymes designed to recognize and break down these synthetic polyester chains efficiently. Instead, they degrade through a passive chemical process called hydrolysis. Water molecules naturally present in body tissues penetrate the suture and attack the ester bonds, snapping the long polymer chains into shorter, soluble fragments. Once the chains are broken into small enough pieces, the body can metabolize these byproducts via normal cellular pathways. Moreover, absorbable suture materials cause reduced local tissue inflammatory responses compared to their non-absorbable counterparts, thereby improving biocompatibility and supporting favorable conditions for tissue regeneration and wound healing [30].

Non-absorbable sutures are fabricated from non-degradable biomaterials, including silk, nylon, polyester, and various metallic alloys. These materials exhibit superior mechanical durability and can maintain long-term tensile strength, making them suitable for applications requiring prolonged mechanical support. Non-absorbable sutures are commonly used to close external skin wounds, facial tissues, and tendons, where sustained high tensile strength is critical for ensuring structural integrity during the healing process [31]. However, despite their advantageous mechanical performance and dimensional stability, a primary limitation of non-absorbable sutures lies in the necessity for a secondary surgical intervention to facilitate their removal once adequate tissue repair has been achieved. This additional procedure increases patient discomfort and introduces a potential risk of postoperative complications, including infection or delayed wound healing [26].

4.2 Materials origin classification

Based on their origin, suture materials are grouped into natural or synthetic, which are further subdivided into absorbable or non-absorbable. Figure 2 shows commercial absorbable and non-absorbable suture materials [25], while Table 1 shows their architectures and brand names. Natural absorbable sutures, including catgut, are primarily composed of highly purified collagen (over 99%) derived from the submucosal layers of sheep or goat intestines. Commercially, catgut is available in two main variants: plain catgut and chromic catgut, the latter being treated with chromium trioxide to enhance its mechanical integrity and reduce its rate of degradation [10]. Chromic catgut demonstrates superior tensile properties and a slower resorption profile than plain catgut, in addition to exhibiting a reduced potential for inducing adverse tissue reactions [32]. The degradation of catgut sutures occurs via proteolytic enzymatic activity, and although they provide peak tensile strength within the first four days post-implantation, they typically experience a complete loss of mechanical strength within two weeks [31]. Despite their widespread historical use, catgut sutures are known to provoke pronounced inflammatory responses, particularly during the initial three days following implantation [33].

Table 1. Materials and structural features of commercial sutures [28]

	Brand Name	Materials	Architecture
Absorbable	Dexon	Polyglycolic acid	Monofilament or braided
	Dexon II	Dexon coated with polycaprolate	
	Vicryl	Coated polyglactin 910	Braided
	Vicryl Rapide	Fast-absorbing coated polyglactin 910	
	PDS	Poly (p-dioxanone)	
	Maxon	Polyglyconate	Monofilament
	Caprosyn	Polyglytone P6211	
	Panacryl	Caprolactone/glycolide	Braided
	Monocryl	Poliglecaprone 25	Monofilament
	Phantom Fiber	Poly-4-hydroxybutyrate	Braided
	Plain gut	Plain gut	
	Chromic gut	Chromic gut	Monofilament
	Biosyn	Glycomer 631	
	Quill ^{m} SRS	Poly(p-dioxanone)	
	Monoderm	Poliglecaprone 25	Barbed
	Ethibond	Polypropylene	
	Ethilon	Aliphatic polymers, nylon 6 and nylon 6,6	Monofilament
	Fiber wire	Polyethylene core with a braided jacket of polyester	
	Force Fiber		
	HiFi	Polyethylene	Braided
	MagnumWire		
	MaxBraid		
Non-absorbable	Prolene	Polypropylene	Monofilament
	TiCron	Polyester	
	UltraBraid	Polyethylene	Braided
	Perma hand	Silk	
	Surgipro	Polypropylene	
	Novafil	Polybutester	Monofilament
	Monosof	Nylon	

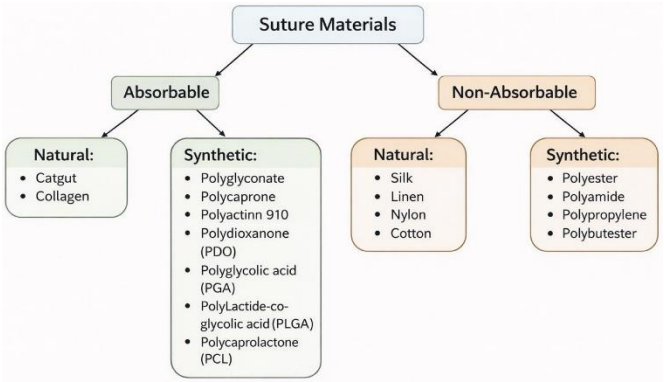

Figure 2. Commercialized suture materials
Adapted from Deng et al. [25]

Furthermore, the absorption kinetics of catgut lag behind its mechanical degradation, often leading to premature tensile failure before sufficient wound healing is achieved, which increases the risk of postoperative complications [34, 35]. Regenerated collagen (RC) has emerged as an alternative biomaterial, offering low immunogenicity and favorable biocompatibility. RC sutures, particularly those fabricated from bovine flexor tendons, have found niche applications in microsurgical procedures. RC can be produced via enzymatic hydrolysis of native collagen-rich tissues or salt extraction methods [31]. To improve its spinnability and mechanical performance, RC is often subjected to chemical modifications, including crosslinking, grafting, or blending with other polymers [36, 37]. Further enhancements in the mechanical strength and thermal stability of RC fibers can be achieved through controlled incubation and maturation processes [31]. For instance, RC derived from bovine dermis using a wet-spinning technique has demonstrated enhanced cell adhesion and proliferation, making it suitable for bioresorbable suture applications [38]. Traditionally utilized as a natural, non-absorbable suture material, silk is processed in braided form, dyed black for surgical visibility, and coated with silicone, wax, or oil to improve handling and reduce tissue drag. Despite these treatments, silk sutures have lower tensile strength than most modern synthetic sutures [39]. Silk's lower tensile strength is not a design flaw but a consequence of its origin as a natural material. Modern synthetic sutures are the product of decades of material science engineering, specifically designed to possess performance characteristics suitable for surgical use. Additionally, the braided structure of silk sutures can facilitate bacterial infiltration, increasing the risk of wound infection [33]. Nevertheless, silk performs well in serum environments and demonstrates water repellency, which can help minimize local inflammation when appropriately managed, although it lacks intrinsic antimicrobial features [34].

Linen sutures, typically manufactured from flax fibers in a twisted multifilament configuration, may also be coated with silicone or polyvinyl compounds to enhance their performance [40]. Linen sutures maintain their mechanical strength following implantation and can increase their tensile capacity by 10% to 20% in hydrated environments [41]. This feature enables them to withstand high tensile loads, making them suitable for securing wound closures under stress [42]. Nylon sutures, often produced in monofilament form, minimize tissue reactivity and reduce the risk of bacterial contamination, making them appropriate for cutaneous wound closure [43]. Nylon offers reliable tensile strength, providing adequate wound support during the healing phase. Commercial nylon suture products, including Ethilon®, are commonly available in black to enhance intraoperative visibility. However, nylon exhibits high shape memory, which complicates knot security,

as the material tends to revert to its original configuration, potentially compromising knot integrity [44].

Cotton sutures, derived from cotton plant fibers, are typically coated with wax to facilitate surgical application. These sutures exhibit a gradual loss of tensile strength, with a 50% reduction typically occurring within six months and complete degradation over two years [43]. Like linen, cotton sutures demonstrate a modest increase in tensile strength when hydrated, typically gaining around 10%. However, their high capillarity increases the risk of bacterial ingress, frequently leading to tissue irritation and infection [34]. Additionally, cotton sutures can generate static electricity, causing adherence to surgical drapes and complicating intraoperative handling [45]. Consequently, the clinical application of cotton sutures is now mainly limited to specific, less critical contexts [25].

Polyglactin 910, commercially known as Vicryl®, is a synthetic, braided, absorbable suture composed of a copolymer of glycolic acid and lactic acid in a 90:10 molar ratio. This multifilament suture is commonly coated with calcium stearate to reduce tissue drag and improve handling characteristics. Vicryl® has high tensile strength up to 14 days after implantation, then hydrolyzes and resorbs within 100 to 120 days. Recent advances in covering Vicryl® sutures with bioactive substances have reduced wound infection and tissue stress during insertion [34]. Vicryl® Rapide, a modified Polyglactin 910, degrades more quickly, making it suited for short-term wound support. Vicryl® Rapide degrades faster because it is chemically modified to be more porous and has a lower molecular weight (short chains). These shorter polymer chains are more easily accessed and broken down by the body's hydrolytic enzymes and water. The process creates a more porous structure, allowing bodily fluids to penetrate the suture more rapidly and accelerate hydrolysis. The result is a suture that loses its strength in 10–14 days instead of the 2–3 weeks for standard Vicryl, making it ideal for superficial, low-tension wounds where long-term support isn't needed (e.g., mucosal repairs, skin closures).

Panacryl®, another derivative with a different monomer ratio, degrades slowly, increasing the suture's mechanical usefulness in vivo [36]. Poly(p-dioxanone) (PDO) is a monofilament absorbable suture recognized for its excellent tensile strength retention and superior knot security. However, its relatively low melting point presents challenges during fiber extrusion, with an increased risk of thermal degradation through pyrolysis [27].

Compared to other synthetic absorbable sutures, including Poly(glycolic acid) (Dexon®) and Poly(glycolide-co-lactide) (Vicryl®), PDO induces a reduced inflammatory response [46]. Commercial monofilament PDO sutures, including PDS® II and PDM®, maintain more than 50% of their original tensile strength after four weeks of implantation [47]. PDO is also fabricated as a barbed suture, exemplified by Quill™ SRS, a bidirectional barbed suture designed to approximate deeper tissues in procedures including abdominal closures and pediatric cardiac surgeries [48].

Polyglyconate sutures, available commercially as Maxon®, are monofilament absorbable sutures composed of glycolic acid and trimethylene carbonate (1,3-dioxan-2-one). These sutures are sterilized using ethylene oxide, which minimizes the risk of eliciting an adverse immune response [36]. Polyglyconate degrades through hydrolysis, initiating around 60 days post-implantation, with complete resorption occurring over approximately 180 days. Known for its superior knot security, Maxon® sutures are extensively utilized in soft tissue approximation, including applications in cardiovascular and peripheral vascular surgeries [34]. This is because Maxon offers the strength and durability needed for vascular structures, along with the handling and safety profile of a superior monofilament suture.

PGA: The simplest aliphatic polyester utilized in absorbable sutures is available in monofilament, braided, and coated forms. PGA degrades exclusively via hydrolysis, beginning with the scission of ester bonds in amorphous regions, followed by the breakdown of crystalline structures [31]. The degradation byproducts are biocompatible and eliminated through renal excretion. After implantation, PGA sutures lose approximately 80% of their tensile strength within 14 days, and degradation accelerates under alkaline conditions, compromising knot integrity [35]. Hydrolytic degradation of PGA can be facilitated by enzymatic action from esterase, trypsin, and chymotrypsin [49]. PGA sutures are frequently utilized for contaminated wound closures due to their rapid absorption profile and predictable degradation [45].

PCL is a versatile, biocompatible, synthetic, absorbable polymer with applications in antibiotic delivery systems and suture fabrication [50]. Prior to processing, PCL must be dehydrated at 40°C for 24 hours to minimize premature hydrolytic degradation [51]. PCL is often copolymerized with PGA to form poliglecaprone, a monofilament suture exhibiting high tensile strength, low tissue drag, and reliable knot security [52]. PCL/PGA sutures are extensively utilized in soft tissue repair, ligations, and aesthetic surgical applications. Indeed, they provide the long-term strength needed for slow-healing tissues and the predictable, complete absorption required to avoid permanent implants, making them ideal for these applications [53].

Poliglecaprone, marketed as Monocryl®, consists of a copolymer of glycolide and epsilon-caprolactone. These absorbable monofilament sutures demonstrate a rapid loss of mechanical strength, with a 50% tensile strength loss within 7 days and complete degradation by 21 days post-implantation [25]. Monocryl® offers excellent knot security and handling characteristics, attributed to its low memory and optimal pliability [34]. It also demonstrates lower microbial adherence compared to non-absorbable monofilament and multifilament sutures, making it suitable for facial, ear, and abdominal wound closures where minimizing hypertrophic scarring is desirable [54].

PLA is a biodegradable, biocompatible polymer that undergoes degradation via hydrolysis and enzymatic activity [15]. Its low modulus and small suture diameter result in minimal mechanical stress on surrounding tissues, making PLA an ideal candidate for soft tissue applications [55]. FDA-approved PLA formulations are widely utilized in biomedical devices and have been enhanced with antimicrobial agents to improve wound healing efficacy [56].

PLGA is a thermoplastic copolymer widely used in absorbable sutures due to its thermal stability and predictable degradation into non-toxic byproducts, lactic acid, and glycolic acid [53]. PLGA sutures have been successfully implemented in bone and tissue engineering, often eliminating the need for secondary surgical removal [57].

Polyester sutures, typically composed of polyethylene terephthalate (PET), are non-absorbable, braided multifilament sutures often coated with polybutylate, Teflon, or silicone to reduce tissue drag [34]. These sutures exhibit exceptionally high tensile strength, second only to metallic

sutures, with negligible loss of mechanical integrity over time [58]. They are extensively utilized in cardiovascular and ophthalmic procedures requiring prolonged mechanical support and minimal tissue reactivity [25, 39].

Polyamide sutures, the first synthetic sutures developed, are available in monofilament and multifilament forms. The multifilament variant is commonly used in dermatological surgeries due to its excellent handling features, though it has been associated with higher tissue reactivity [59]. Monofilament polyamide sutures, including nylon, demonstrate long-term tensile retention, losing approximately 30% of their tensile strength over two years, whereas the multifilament forms degrade entirely within six months [34].

Polypropylene, produced through the catalytic polymerization of propylene, is a monofilament suture frequently employed in cutaneous wound closures [34]. Due to its smooth surface finish, it offers consistent tensile strength post-implantation, is easy to handle, and elicits a minimal inflammatory response. However, its low-friction surface can compromise knot security, necessitating meticulous surgical technique during closure [39]. Polybutester sutures, consisting of polybutylene, polyglycol, and polytetramethylene terephthalate, are monofilament sutures with thermoplastic features [34]. These sutures offer superior handling, minimal memory, and enhanced knot security. Furthermore, polyester sutures are associated with a lower risk of hypertrophic scarring than other monofilament materials, including nylon [44]. Their mechanical reliability and biocompatibility make them suitable for soft tissue approximation and general surgical applications [39].

4.3 Yarn construction classification

Based on yarn structures (structural features), sutures can be classified as monofilament, multifilament, and braided sutures [48], as shown in Figure 3. Monofilament sutures are composed of a single, continuous filament, while multifilament sutures are manufactured by twisting or braiding multiple individual fibers into a unified strand. In general, multifilament sutures exhibit superior handling characteristics, enhanced pliability, and improved knot security compared to monofilament sutures. The increased surface friction and flexibility of multifilament structures contribute to maintaining secure knots with less risk of slippage.

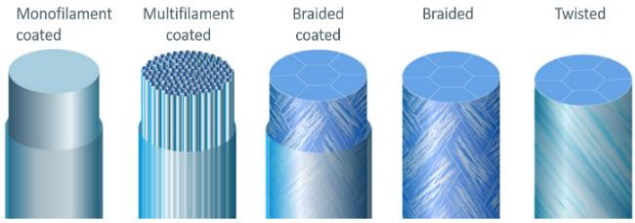

Figure 3. Suture classification based on their feature structure [48]

Conversely, monofilament sutures tend to demonstrate lower knot-holding capacity due to their smooth surface topology and higher intrinsic stiffness. They are more susceptible to knot failure or breakage under mechanical stress [27]. The difference comes down to surface structure and friction, as multifilament sutures are braided or twisted from many fine fibers. This creates a rough, textured surface, making them softer and more flexible, like a piece of yarn, so they are easier to tie and manipulate. The high friction between the braided strands causes them to "grip" each other tightly, preventing knots from slipping loose. On the other hand, monofilament sutures are a single, solid strand. This creates a smooth, slick surface, making them harder to handle and tie. Their smooth surfaces slide against each other easily, so knots are more prone to slipping and require more "throws" to secure.

Moreover, braided (multifilament) sutures possess elevated capillarity, facilitating fluid wicking along the fiber structure. This phenomenon increases the risk of fluid infiltration, creating a potential pathway for bacterial migration, thereby elevating the likelihood of tissue inflammation or immune responses compared to monofilament sutures [26, 60]. Despite these concerns, the larger surface area provided by multifilament constructs enhances their capacity for surface coatings, including antimicrobial agents, which can mitigate the risk of infection and improve overall biological performance. The structural flexibility and increased surface reactivity of multifilament sutures make them particularly advantageous for applications requiring antimicrobial functionality [28]. Braided sutures, illustrated in Figure 3, are specialized surgical sutures engineered with projections or barbs along their length, facilitating tissue approximation without the need for traditional knot tying. These sutures are available in unidirectional and bidirectional configurations. Unidirectional barbed sutures are typically fabricated from monofilament fibers, with barbs oriented in a single direction along the suture body. One end of the suture is swaged to a surgical needle, while the opposite end is secured by an anchor mechanism or a terminal knot to prevent retrograde movement. In contrast, bidirectional barbed sutures, also produced from monofilament structures, feature two sets of barbs arranged in opposing directions, typically originating from the midpoint of the suture. Each end of the suture is swaged to a needle, facilitating simultaneous tissue engagement from the central anchoring point outward. Barbed sutures offer distinct advantages over conventional monofilament and multifilament sutures. Their knotless design enables secure closure of multiple tissue layers, improving load distribution and maintaining consistent wound tension, which can enhance the overall tensile strength of the wound closure [61]. Furthermore, the barb geometry reduces suture slippage. It minimizes the potential for tissue strangulation while inhibiting bacterial adherence and colonization, thereby decreasing the risk of surgical site infections and inflammatory responses [26]. Despite these clinical advantages, barbed sutures present certain limitations. The sharp barbed tips can inadvertently perforate surgical gloves, posing a risk of cross-contamination between the surgical team and the patient, potentially leading to nosocomial infections [28]. Additionally, the mechanical process of barb formation, typically involving cutting or scoring the suture's outer layer in a helical pattern, may compromise the core integrity of the suture, reducing the cross-sectional area and subsequently lowering the tensile strength of the suture material. Nevertheless, with appropriate surgical handling techniques, barbed sutures have demonstrated reliable antimicrobial features and have effectively promoted optimal wound healing, particularly in dermal and subcutaneous tissue closures [25].

Braided yarns are easy to work with but prone to contamination. The reason lies entirely in their surface structure, which is composed of many fine, intertwined fibers.

Braided sutures are inherently soft and flexible, making them easy for a surgeon to bend, loop, and manipulate. This pliability gives them a superior "hand," meaning they are less stiff and do not have the "memory" (spring-back tendency) that monofilaments have. They lie flat and are easier to tie. The microscopic crevices and rough texture of the braid create immense internal friction. When a knot is tied, the strands grip each other tightly, preventing the knot from slipping or loosening easily.

The tiny interfilamentous spaces act like a network of microscopic straws. If bacteria or fluid are present, they can be wicked deep into the core of the suture, away from the body's immune defenses and circulating antibiotics. Additionally, a braided suture has a vastly larger total surface area compared to a smooth monofilament of the same diameter. This provides more "real estate" for bacteria to adhere to and colonize. Based on these findings, a surgeon chooses a braided polyester suture when high strength, permanence, and easy knot-tying in a clean site are needed. They choose a monofilament polypropylene suture when minimal tissue reaction and low infection risk are required, accepting the challenge of tying it securely. The structure and material are chosen together to meet the specific clinical need.

4.4 Specified sutures

Specialized surgical sutures have been designed for different wound-healing circumstances, such as antimicrobial and smart sutures for specific clinical uses and performance.

The term "suture smartness" refers to a suture's capacity to revert to its original shape after deformation. These sutures may be elongated at temperatures below the critical threshold before implantation. Initially, the suture is placed loosely on the wound site; subsequently, when the temperature increases, either due to physiological factors or external causes, the suture reverts to its original configuration.

Intelligent sutures have excellent flexibility and mechanical properties for attaining self-tightening capabilities [28]. A tight knot during knot tying might damage healthy cells, resulting in skin necrosis. A poorly sealed wound line would allow extraneous substances to penetrate the wound site. Consequently, a self-tightening smart suture that operates at body temperature is highly desirable.

Antimicrobial sutures have improved surgical site infection prevention, while smart sutures may monitor physiological parameters or administer therapeutic drugs in real time.

5. TECHNIQUES OF LOADING SURGICAL SUTURES WITH DRUGS

Loading surgical sutures with drugs transforms them from passive wound-closing devices into active therapeutic platforms, often called "drug-eluting sutures" or "multifunctional sutures." The drug must remain stable during the loading process (e.g., high heat in melt-spinning) and throughout its shelf life.

The techniques can be broadly categorized into physical and chemical methods. The choice depends on the drug's properties (hydrophilicity, molecular weight, stability), the suture material (absorbable, like PLGA or PCL, or non-absorbable, like nylon or silk), and the desired release profile [62].

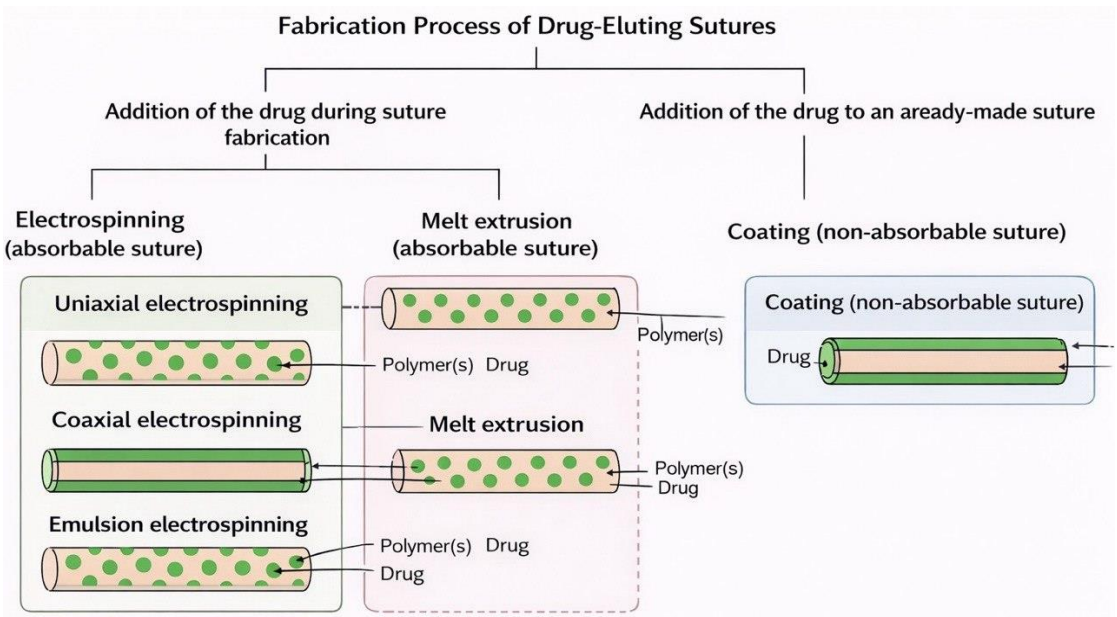

Figure 4. The manufacturing process of drug-eluting sutures
Adapted from Xu et al. [63], published by Wiley

Physical loading/coating techniques

These are the most common and commercially explored methods:

•Dip-coating method process: The suture is dipped into a solution or dispersion containing the drug and a polymer binder (e.g., chitosan, gelatin, PLGA). It is then dried, leaving a thin drug-loaded coating on the surface. The mechanism involves drug release via diffusion from the coating and/or erosion of the polymer matrix. It is a simple, cost-effective, and scalable method, but it suffers from burst release (a large initial dose) being common, and the coating may wear off during handling or knotting.

•Electrospinning process: A high voltage is applied to a polymer solution containing the drug, creating nanofibers that

are directly deposited onto the suture surface, forming a core-sheath structure (the suture is the core, and the nanofibers are the drug-loaded sheath). During this process, controlled release through diffusion from the nanofibrous mat occurs. This process has a high surface area, allowing for high drug loading and tunable release kinetics, but its disadvantages include a more complex setup and the potential for solvent residues.

•Soaking/impregnation process: The suture is soaked in a concentrated drug solution for an extended period, allowing the drug to diffuse into the pores or swellable regions of the polymer. The mechanism is that drug release is based on diffusion out of the suture matrix. It is very simple and works best with absorbable, porous sutures [63].

Figure 4 shows the manufacturing process of drug-elution sutures [63].

6. DISCUSSION

The development of surgical sutures is a continuous process of balancing competing material properties to achieve optimal clinical outcomes. Our analysis reveals several fundamental trade-offs that define the field.

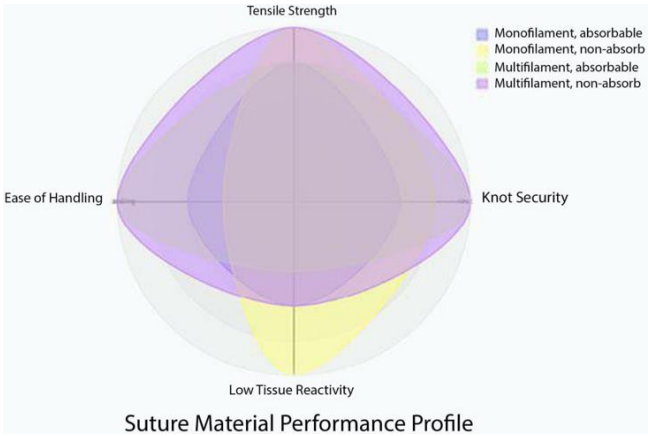

Figure 5. Radar chart comparing key performance parameters (e.g., tensile strength, knot security, tissue reactivity, handling) across monofilament vs. multifilament and absorbable vs. non-absorbable sutures

First, the conflict between operability and anti-infectivity is epitomized by the choice between braided and monofilament sutures. While braided structures offer superior handling and knot security, their multifilament architecture creates a textured surface and capillary action that can harbor bacteria, increasing the risk of infection. Conversely, the smooth surface of monofilament sutures minimizes bacterial adhesion and tissue drag but often results in poor knot security and challenging handling, presenting a direct compromise for the surgeon.

Second, a central challenge lies in precisely matching a suture's degradation rate with the wound's healing speed. An absorbable suture that degrades too quickly risks wound dehiscence due to premature loss of mechanical strength. Conversely, a suture that persists too long can provoke a chronic inflammatory response or impede the healing process by acting as a foreign body. This necessitates a deep understanding of tissue-specific healing timelines to engineer polymers with predictable, tailored absorption profiles. Finally, the pursuit of added functionality, such as antibacterial properties or drug delivery, introduces potential negative impacts on the material's intrinsic properties. Incorporating antimicrobial agents or drugs can alter the suture's surface chemistry, mechanical strength, and degradation kinetics. For instance, the burst release of an antibiotic may weaken the polymer matrix, while the additives themselves can cause local tissue toxicity or inflammatory reactions. Therefore, any functional enhancement must be carefully evaluated against its potential to compromise the suture's primary mechanical and biocompatible roles.

The radar chart (Figure 5) shows the following findings:

1. Multifilament sutures (blue & purple) dominate in knot security and ease of handling. Their braided structure is easy to tie, but often has lower tissue reactivity scores.

2. Monofilament, non-absorbable (light yellow) stands out for low tissue reactivity and permanent tensile strength, but is the most difficult to handle and tie.

3. Monofilament, absorbable (light blue) is a true "middle-of-the-road" option, with no standout highs or lows, offering a balance of properties for internal soft tissue closure where prolonged support isn't needed.

This visualization makes the key trade-offs between the different suture types immediately obvious.

7. CONCLUSIONS

This review is important because it demonstrates that a suture is an active implant, not merely a passive thread. Optimizing its polymeric design is fundamental to improving surgical outcomes, reducing complications, and advancing regenerative medicine. It underscores that suture properties—from tensile strength to degradation rate—are not arbitrary but are direct consequences of polymer chemistry and manufacturing. The evolution from natural materials like silk to a diverse portfolio of synthetic polymers (PGA, PLA, PCL, and their copolymers) represents a triumph of materials science, enabling precise control over suture performance in the body. The implications of this review are significant, especially for materials scientists who must move beyond simple monofilaments and braids. The future lies in advanced engineering approaches such as developing multi-component core–sheath structures, incorporating biomimetic coatings, and creating drug-eluting “smart” sutures that actively promote healing and prevent infection. For surgeons, understanding this structure-property relationship is crucial for evidence-based device selection. It empowers surgeons to choose not just a suture, but a predictable healing partner—matching a suture's engineered degradation profile and strength retention to the specific biological healing timeline of a tissue (e.g., fast for skin, slow for fascia). Finally, for the industry, the focus must be on developing next-generation platforms that offer superior performance and address unmet clinical needs, such as sutures for compromised tissues (diabetic wounds, irradiated fields) or with enhanced antimicrobial properties.

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