SUTURES

1. Introduction

Wound closure biomaterials are generally divided into three major categories: sutures, staplers/ligating clips, and tissue adhesives. The sutures have the longest history, received the most attention, and are the most widely used in wound closure. Ligating clips and staplers facilitate anastomosis with minimal trauma, necrosis, or interruption of tissue function, and their use has steadily increased in specific clinical conditions, particularly the availability of synthetic absorbable ligating clips and staplers. Tissue adhesives are the least frequently used for wound closure at the present time, even though they received considerable attention in the 1960s; however, some new tissue adhesives have recently received increasing attention. There are several reviews about wound closure biomaterials (1–9). In this article, the focus is on suture-based wound closure biomaterials because they are the most frequently used and studied. All aspects of suture-based wound closure biomaterials, their classification, and chemical, physical, mechanical, biological, and biodegradation properties are concisely covered.

A *suture* is a strand of material, either natural or synthetic, used to ligate blood vessels and to approximate tissue together. Suture materials are the earliest and most frequent application of textile materials for surgical wound closure. Linen was used as a suture material as early as 4000 years ago. Since then, numerous materials have been used as ligatures and sutures: iron wire, gold, silver, dried gut, horse hair, strips of hide, bark fibers, silk, linen, and tendon. Among these catgut and silk dominated the suture market until 1930. The introduction of steel wire and synthetic nonabsorbable fibers such as nylon, polyester, and polypropylene during and after World War II greatly expanded the chemical composition of suture materials. During the early 1970s, the introduction of two synthetic absorbable suture materials, Dexon (Davis & Geck, Danby, Conn.) and Vicryl (Ethicon, Inc., Somerville, N.J.) opened a new milestone for suture materials. Owing to their precisely controlled manufacturing processes and uniform and reproducible properties, these synthetic absorbable sutures have received a great deal of attention from both surgeons and researchers. Since then, several new synthetic absorbable suture materials such as PDSII and Panacryl (Ethicon), Maxon (US Surgical/Davis & Geck), Monocryl (Ethicon), and Biosyn (US Surgical, Norwalk, Conn.) have been commercially available. The most important advantage of synthetic absorbable sutures is their reproducible and predictable degradability inside a biological environment. This property will enable the sutures to minimize chronic undesirable tissue reactions after the sutures have lost most of their mechanical properties. The latest property introduced in sutures is multifunctionality, and a typical example is Triclosan-coated Vicryl, Vicryl Plus, which has antimicrobial capability to reduce the chance of wound infection (10,11). Today, surgeons can choose between a large number of suture materials with various chemical, physical, mechanical, and biological properties.

2. Needles

The surgical needle, to which a suture is attached, has the primary function of introducing the suture through the tissues to be brought into apposition. Thereafter, the tissues are maintained in apposition by the suture until physiologic healing of the wound has occurred. Ideally, the needle has no role in wound healing, but inappropriate needle selection can prolong the operating time and/or damage tissue integrity leading to such complications as tissue necrosis, wound dehiscence, bleeding, leakage of anastomoses, and poor tissue apposition. For maximum effectiveness, the surgical needle must be able to carry the suture material through tissue with minimal trauma, and to achieve this goal, needles are required to satisfy several criteria. For example, the needle must be sharp to ensure easy and rapid tissue penetration and, on passage through tissue, not cause undue trauma. The latter criterion is achieved, in part, by fabricating needles from corrosion- and abrasion-resistant materials and providing the needle with a smooth, scratch-free surface finish to minimize frictional effects.

Needles are commonly fabricated from stainless steel, a material that has high strength, is readily available, presents few manufacturing problems, can be polished to a smooth finish, and is relatively inexpensive. A wide variety of surgical needles exist, but all types have basically three components: the point, the body, and the eye, swage, or attachment end. Virtually all surgical needles used in modern surgery are swaged, with the suture being bonded to the needle to form a continuous unit. Recently, laser-drilled needles have become increasingly popular and these are manufactured by laser-drilling a hole in the distal end of the needle body parallel to the axis of the needle (12,13). The suture is retained within the hole held by an adhesive. The laser-drilled needle has a smoother outer circumference. The laser-drilled swage needle has the advantage of smaller size differential between needle and suture thread body (the needle to suture thread diameter ratio) than for the traditional crimped or channel needle. This reduced needle to suture thread diameter ratio has been reported to facilitate surgical suturing and aid in wound healing (12).

The most frequently used needles are curved and the curvatures are usually 1/4, 3/8, 1/2 or 5/8 of the arc of a circle; that is, a curvature of 90°, 135°, 180°, and 225°. Curved needles are favored in surgery because they are more predictable in their passage through tissue, but, as a result of the curvature, they require the use of needle holders.

3. Commercial Sutures, Their Sizes, and Physical Configurations

Suture materials are generally classified into two broad categories: absorbable and nonabsorbable. Absorbable suture materials generally lose their entire or most of tensile strength within two to three months; those which retain most of their initial strength longer than two to three months are nonabsorbable. The absorbable suture materials are catgut (collagen sutures derived from sheep intestinal submucosa), reconstituted collagen, polyglycolide (Dexon, Dexon II, Dexon S), poly(glycolide-lactide) random copolymer (Vicryl and Pana-

cryl), antimicrobial-coated Vicryl (Vicryl Plus), poly-*p*-dioxanone (PDS, PDSII), poly(glycolide-trimethylene carbonate) block copolymer (Maxon), poly(glycolide-ε-caprolactone) (Monocryl), and gycolide-dioxanone-trimethylene carbonate block copolymer (Biosyn). The nonabsorbable sutures are divided into the natural fibers (ie, silk, cotton, linen) and synthetic fibers [ie, polyethylene, polypropylene, polyamide, polyester, polytetrafluoroethylene (Gore-Tex), and stainless steel]. Table 1 summarizes most commercial suture materials that are available mainly in the United States and Europe and Pacific, their generic and trade names, physical configurations, and manufacturers.

Suture materials are also classified according to their size. Currently, two standards are used to describe the size of suture materials: USP (United States Pharmacopoeia) and EP (European Pharmacopoeia) (9). The USP standard is more commonly used and the size is represented by a series combination of two Arabic numbers: a zero and any number other than zero, such as 2-0 (or 2/0). The higher the first number, the smaller the suture material is. Sizes greater than 0 are denoted by 1, 2, 3 etc. This standard size also varies with the type of suture material. In the EP standard, the code number ranges from 0.1 to 10. The corresponding minimum diameter (mm) can be easily calculated by taking the code number and dividing by 10. The EP standard does not separate natural from synthetic absorbable sutures as USP does.

In terms of the physical configuration, suture threads can be classified into monofilament, multifilament, twisted, and braided. Suture materials made of nylon, polyester, and stainless steel are available in both multifilament and monofilament forms. Catgut, reconstituted collagen, and cotton are available in twisted multifilament form, while Dexon, Vicryl, Monosyn, Polysorb, PolySyn FA, Safil, BioSorb, , and polyester-based, polyamide-based suture materials are available in the braided multifilament configuration (see FIBERS, POLYESTER). PDS, Maxon, Monocryl, Biosyn, Caprosyn, MonoPlus, polypropylene, and Gore-Tex (polytetrafluoroethylene) suture materials exist in monofilament form only (see Perfluorinated Polymers, Polytetrafluoroethylene). Stainless steel metallic suture materials can be obtained in either monofilament or twisted multifilament configurations. Another unique physical configuration of suture material is available in polyamide (nylon 6) and has the trade name Supramid, which has a twisted core covered by a jacket of the same material (see POLYAMIDES, FIBERS).

Suture materials are frequently coated to facilitate their handling properties, particularly a reduction in tissue drag when passing through the needle tract and the ease of sliding knots down during knotting (ie, knot tie-down). Although nonabsorbable bee wax, paraffin wax, silicone, and polytetrafluoroethylene (Teflon) are the traditional coating materials, new coating materials have been reported, particularly those that are absorbable (14–16). There are basically two types of absorbable coating materials: water-soluble and -insoluble. Water-insoluble coating materials have similar chemical constituents to the suture and they are broken down by hydrolysis. They remain on the suture surface longer than water-soluble coatings. A typical example is polyglactin 370 used for Vicryl suture. Dexon II sutures have a polycaprolate coating which is water-insoluble. Water-soluble coating materials dissolve promptly to reveal the uncoated suture underneath after wound closure. A typical example is polox-

amer 188 found on Dexon Plus. Multifilament sutures are more commonly coated than monofilament sutures. For example, multifilament Vicryl and Dexon Plus or II have coating materials applied, while monofilament PDS and Maxon sutures have no coatings.

Although coating of suture materials facilitates easy passage through tissue, it frequently results in poor knot security. For example, Dexon Plus and coated Vicryl require four or five square throws to form secure square knots, while the uncoated Dexon and Vicryl sutures form secure knots with only 2 throws (1 = 1) (17,18).

There are several other patented procedures and materials reported to improve either *knot tie-down* performance (the ease of sliding a knot down the suture into place during knotting) or/and *knot security* (the ability of a knot to hold after knotting) (19–22). In general, a coating designed to improve knot tie-down would reduce knot security. It is difficult to achieve both ease of knot tie-down and enhanced knot security of sutures. There are very few reported treatments that would achieve these two contradictory and mutually exclusive properties (ie, ease of knot tie-down and enhanced knot security). One of them is the use of a combination of both coating and textured yarns (19). Other recently reported absorbable but water-insoluble coating materials that could improve knot tie-down and knot security are high molecular weight poly-ε-caprolactone, copolymer of at least 90% by weight of caprolactone and 10% at most of other biodegradable monomers such as glycolide, lactide, and their derivatives (20,21), or a random copolymer of 25–75% by weight of glycolide and the remaining trimethylene carbonate (22).

There are four essential properties of suture materials: physical and mechanical properties, handling properties, biological properties (biocompatibility), and biodegradation properties. Table 2 summarizes the characteristics of each of the four essential properties. These characteristics are interrelated. For example, capillarity of a suture material under physical/mechanical properties is closely related to the ability of the suture to transport bacteria, which is a biological property. The modulus of elasticity under physical/mechanical properties is frequently used to relate to pliability of sutures under handling property. Descriptions of each of these essential properties are given in the following sections.

4. Physical and Mechanical Properties

Physical and mechanical properties are probably most important in terms of suture function; ie, close wounds and carry physiologic load during healing. These properties include those related to strength, stiffness, viscoelasticity, coefficient of friction, compliance, size, form (monofilament or multifilament), fluid absorption and transport, etc. Strength includes knotted and unknotted (straight pull) tensile strength, modulus of elasticity (relating to stiffness), elongation at break, and toughness.

Tensile strength is the most frequently reported and studied physical/mechanical property of suture materials. A larger-size suture has a higher tensile breaking force than the same suture of a smaller size, even though the two

sutures may have the same tensile strength. Therefore, a meaningful comparison of tensile breaking force of several sutures should be done under same suture size (diameter) and form. In addition, knotted tensile strength or breaking force is frequently lower than that for the unknotted suture. Strength values are obtained in either dry or wet conditions. Among these physical and mechanical properties, viscoelasticity, bending stiffness, and compliance are the least studied and understood.

Bending stiffness closely relates to handling characteristic of suture materials, particularly knot security. There are two reported studies of bending stiffness of sutures (23,24). One study (23) of bending stiffness was based on the force required to bend a suture to a predetermined angle. The measured bending force was converted to flexural stiffness in lb/in.^{2} according to an ASTM formula. Braided sutures are generally more flexible than monofilament sutures of equivalent size, irrespective of their chemical constituents. Coated sutures have a significantly higher bending stiffness than the corresponding uncoated ones. This increase in bending stiffness is attributable to the loss of mobility of constituent fibers under bending force. An increase in suture size significantly increases their stiffness, and the magnitude of increase depends on the chemical constituent of the suture. The large porous volume inherent in Gore-Tex monofilament suture is the reason for its lowest flexural stiffness. In a second reported bending stiffness study of a few sutures (24), a constant weight was attached to each of the two ends of a bended suture and the distance between these two ends was measured after 1-min loading. Bending stiffness data from the two cited approaches (23,24) generally agree with each other; ie, braided sutures are generally more flexible than monofilaments, and Gore-Tex suture has the lowest bending stiffness.

Suture compliance is a mechanical property that closely relates to the ease of a suture to elongate under a tensile force. It is believed that the level of suture compliance should contribute to the compliance of tissues at the anastomotic site. Suture compliance is particularly important in surgery where there is a tubular anastomosis, such as vascular anastomoses. Compliance mismatch between a vascular graft and host tissue has long been suggested as one of the several factors contributing to graft failure (25). Since sutures are the only foreign materials in the anastomotic site, it is expected that a wide range of suture compliance might result in different levels of anastomotic compliance. There is only one reported study that examined the effect of suture compliance on the compliance of arterial anastomotic tissues closed with two suture materials vastly different in suture compliance: 6/0 Novafil and Prolene (26). Novafil is an elastomeric suture made from polybutester and is characterized by a high elongation at low tensile force, low modulus of elasticity, and high hystersis, whereas Prolene suture has a relatively higher modulus of elasticity, low elongation at low tensile force, and low hystersis. In a clinical condition of minimal tubular compliance and diameter mismatch such as artery–artery anastomoses, a far more compliant anastomosis was achieved with Novafil ($5.9 \pm 2.0\%$) than with Prolene ($3.3 \pm 0.6\%$) suture. Thus, arterial anastomoses closed with a more compliant suture produced arterial anastomotic compliance on average over 75% more than those closed with a less compliant suture.

5. Handling Property

Handling property describes the "feel" of suture materials by surgeons during wound closure. It is the only category of suture properties that is difficult to evaluate objectively. Handling property includes pliability (or stiffness), ease of knot tie-down, knot security, packaging memory, surface friction, viscoelasticity, tissue drag, etc. These are directly and indirectly related to physical/mechanical properties of a suture. For example, the term *pliability* of a suture is a subjective description of how easily a person could bend it, and hence relates to the surgeon's feel of a suture during knot tying. It is directly related to the bending modulus of a suture and indirectly to coefficient of friction. Packaging memory, another handling property that indirectly relates to pliability, is the ability to retain the kink form of sutures after unpacking them. The ability to retain such kink form after unpacking would make surgeons' handling of sutures more difficult during wound closure, particularly tying a knot. This is because sutures with high memory, such as nylon, polypropylene, PDS, and Maxon, tend to untie their knots as they try to return to their kink form from packaging. Thus, packaging memory should be as low as possible. In general, monofilament sutures have more packaging memory than braided ones. The three exceptions are the newly available Monocryl, Biosyn, and Gore-Tex monofilament sutures, which were reported to have exceptionally low packaging memory.

Knot tie-down and security describe how easily a surgeon can slide a knot down to the wound edge and how well the knot will stay in position without untying or slippage. This handling property relates to surface and mechanical properties of sutures. A relatively smooth surface like monofilament or coated braided suture would have a better knot tie-down than a suture with a rough surface such as an uncoated braided suture, if everything else is equal. The coefficient of friction of sutures also relates to knot tie-down and security. A linear relationship between knot security and coefficient of friction has been reported (27). A high coefficient of friction would make knot tie-down difficult but would lead to a more secure knot. This is because a high friction suture could provide additional frictional force to hold the knot together. This high friction suture surface also makes the passage of suture strands difficult during knot tie-down. It thus appears that knot tie-down and knot security are two contradictory requirements. A method to objectively quantify the knot tie-down capacity of 2/0 silk, polyester sutures, Gore-Tex, and an experimental ultrahigh molecular weight polyethylene suture (Nesplon) has been reported (24).

6. Biological Property

Biocompatibility of suture materials describes how sutures, which are foreign materials to the body, could affect surrounding tissues and how the surrounding tissues could affect the properties of sutures. Thus, biocompatibility is a two-way relationship. The extent of tissue reactions to sutures depends largely on the chemical nature of sutures and their degradation products if they are absorbable. Sutures from natural sources such as catgut and silk usually provoke more

tissue reactions than synthetic ones because of the availability of enzymes to react with natural biopolymers. Besides the most important chemical factors, physical form and the amount and stiffness of suture materials have been reported to elicit different levels of tissue reactions. For example, a stiff suture would result in stiff projecting ends in a knot where cut. These stiff ends could irritate surrounding tissues through mechanical means, a problem associated with some monofilament sutures but generally not found in braided multifilament sutures.

Because the quantity of a buried suture relates to the extent of tissue reaction, it is a well-known practice in surgery that one should use as little suture material as possible, such as a smaller knot or a smaller size, to close wounds. The use of a smaller size of suture for wound closure without detriment to the provision of adequate support to wounds and cutting through wound tissue is due to the square relationship between diameter and volume which suggests that a slight increase in suture size or diameter would increase its volume considerably.

There are two basic means to study biocompatibility of suture materials: cellular response and enzyme histochemistry. The former is the most frequently used and provides information about the type and density of inflammatory cells at a suture site. In the cellular response approach, sutures without tension are implanted in the gluteal muscle of small animals like rats. This implantation site has given a very consistent reproducible cellular response for valid comparisons, even though it is not a common site for suture in surgery. However, use of this common test procedure has been questioned, particularly in orthopedic surgery (28) because of the observed inflamed nature of the postoperative synovial tissue and the mechanically stressed nature of the suture. Histological stains with a variety of dyes such as the most frequently used H & E are the standard methods of evaluation of cellular activity at the suture sites. Figure 1 is a typical example of histological photomicrographs of PDS and Maxon sutures at 35 days postimplantation in a variety of tissues (29). In addition to a qualitative description of cellular activities, tissue response could also be graded by the most frequently used and accepted Sewell and co-workers method or its modification (30).

The enzyme histochemical approach is a more objective, quantitative, consistent, and reproducible method than cellular response. Enzyme histochemistry is based on the fact that any cellular response to a foreign material is always associated with the presence of a variety of enzymes; however, this approach is more tedious and requires more sophisticated facilities and better experience. The data obtained provide additional insight into the functions of those cells appearing during various stages of wound healing. The enzymatic activity of a suture implant site is quantified by microscopic photometry of a cryostat section of the tissue. For example, the high level of cellular response to silk suture observed in a histological study has been confirmed by an enzyme histochemical study (31–33). Enzyme histochemistry is also useful for studying the biodegradation mechanism of absorbable sutures because not only natural absorbable sutures are degraded through the enzymatic route but also the degradation products must be metabolized via enzyme activity.

The normal tissue reaction to sutures can be viewed in three stages, according to the time for the appearance of a variety of inflammatory cells (1,31,33,34).

They are as follows initial infiltration of polymorphonuclear leukocytes, lymphocytes, and monocytes during the first 3–4 days (ie, acute response); appearance of macrophages and fibroblasts from day 4 to day 7; and beginning maturation of fibrous connective tissue formation with chronic inflammation after the 7th to the 10th day. During the first 7 days post-implantation, there is virtually no difference in normal tissue reaction between synthetic absorbable and nonabsorbable sutures. However, a slightly higher inflammatory reaction to synthetic absorbable sutures could persist for an extended period until they are completely absorbed and metabolized, while synthetic nonabsorbable sutures, in general, are characterized with a minimal chronic inflammatory reaction with a thin fibrous connective tissue capsule surrounding the sutures usually by 28 days post-implantation.

In addition to the normal tissue reactions to sutures, there are several adverse tissue reactions that are suture- and site-specific. Some examples include urinary stone or calci formation, granuloma formation, thrombogenicity, propensity toward wound infection and recurrence of tumor after radical surgery, and allergy.

Monofilament sutures are considered to be a better choice than multifilament ones in closing contaminated wounds. This is because not only do multifilament sutures elicit more tissue reactions which may lessen tissue ability to deal with wound infections but also multifilament sutures have a capillary effect which could transport microorganisms from one region of the wound to another. Multifilament sutures generally elicit more tissue reactions than their monofilament counterparts because inflammatory cells are able to penetrate into the interstitial space within a multifilament suture and invade each filament. Such an invasion by inflammatory cells, well evident in histological pictures, could not occur in monofilament sutures. Thus, the available surface area of a suture to tissue bears a close relationship to the level of tissue reaction that a suture elicits.

7. Biodegradation and Absorption Property

Biodegradation and absorption properties are the most important issue of absorbable sutures, but are far less relevant for most nonabsorbable sutures, particularly those of synthetic nature. This biodegradation property is also responsible for the fact that absorbable sutures do not elicit permanent chronic inflammatory reactions found with nonabsorbable sutures. The most important characteristics in biodegradation and absorption of sutures are the strength and mass loss profiles and biocompatibility of degradation products (see Biodegradable, Polymers, Medical Applications). Although there is a wide range of strength and mass loss profiles among the available absorbable sutures, they have one common characteristic: strength loss always occurs much earlier than mass loss, as shown in Table 3. This suggests that absorbable sutures retain a large portion of their mass in tissue while they have already lost the mechanical properties required to provide support for tissues during wound healing. This discrepancy between the duration for tensile strength loss and the duration for mass loss also applies to the new long-lasting absorbable sutures that have come to the

market more recently, such as Panacryl. Both Panacryl and Vicryl are made from copolymers of glycolide and L-lactide. Vicryl is predominately glycolide (90%), and Panacryl is made predominately from L-lactide (95%). Because of the much slower degradation of the L-lactide component than the glycolide counterpart, Panacryl is expected and was reported to be essentially absorbed in rat tissues between 1.5 and 2.5 years, and 60% and 20% of the original tensile strength of Panacryl remained at the end of 6 month and 2 years post-implantation, respectively, while Vicryl has less than 10% strength remaining at the end of 3 weeks post-implantation. The goal of extending tensile strength retention of absorbable sutures over many month periods has certainly been met by Panacryl suture, but at the expense of extended suture mass remnants over a much longer period than other synthetic absorbable sutures. The line between absorbable and nonabsorbable sutures has become difficult to distinguish as the absorption of this new synthetic absorbable suture in tissues takes years to complete. Whether such a long duration of suture mass retention for absorbable sutures would impose any undesirable tissue reactions remains to be seen. In July 2002, Ethicon decided to discontinue the Panacryl product line globally. An ideal absorbable suture should match mass loss and strength loss profiles, and none of the current commercial absorbable sutures can achieve this ideal biodegradation property.

The observed wide range of strength and mass loss profiles among the available absorbable sutures is attributable not only to the chemical differences among the absorbable sutures but also to a variety of intrinsic and extrinsic factors, such as pH, electrolytes, stress applied, temperature, γ irradiation, superoxide, microorganisms, and tissue type, to name a few. Among these intrinsic and extrinsic factors, the role of superoxide on the biodegradation of absorbable sutures appears to be one of the most interesting factors because of the unusually fast loss of mechanical property and unique surface morphology observed (35). For example, at a 0.005 *M* superoxide ion concentration and room temperature, the 5 synthetic absorbable sutures retained 20–70% of their original tensile breaking force at the end of 24 h, as shown in Figure 2. The bulk of the loss of tensile breaking force of these sutures occurred during the initial 2-h period. Even the most superoxide ion resistant PDS II suture showed an appreciable loss of tensile breaking force at 0.005 *M* superoxide ion concentration. Like PDS II sutures, Dexon, Vicryl, and Maxon sutures all showed most of their loss of tensile breaking forces between 2- and 24-h period. The order of these five absorbable suture materials toward the superoxide ion sensitivity at this relatively higher superoxide ion concentration was the same as the lower superoxide ion concentration case: Monocryl > Maxon > Vicryl > Dexon > PDS II. There is no change in tensile breaking force of these absorbable sutures in regular saline buffer media at 25°C for as long as several days (9).

Upon biodegradation, absorbable sutures have shown quite interesting surface morphology, and some examples are shown in Figure 3. For example, multifilament Dexon sutures that were subjected to γ-irradiation treatment and hydrolytic degradation in buffer solution showed very regular circumferential surface cracks along the longitudinal fiber axis and had the appearance of "corn-like" structure (Figs. 3a & 3b). Upon γ-irradiation treatment and hydrolytic degradation in buffer solution, monofilament Maxon sutures, however,

showed both circumferential and longitudinal surface cracks (Fig. 3c) and the subsequent peeling off these surface cracks (Fig. 3d). The appearance of moon crater-shaped impressions of various sizes (about 10–100 μm diameter) on Monocryl suture at a superoxide ion concentration (>0.005 *M*, Fig. 3e) is unique because such circular impressions were never observed in the hydrolytic degradation of all existing absorbable sutures in a conventional saline buffer medium or *in vivo*. The formation of moon-crater-shaped impressions on Monocryl and Maxon sutures deviates from the conventional understanding of the anisotropic characteristic of fibers. It appears that these impressions started randomly on suture fiber surface and propagated concentrically (ie, uniformly at all angles), irrespective of the fact that all fibers are highly anisotropic. In the reported morphological studies of all existing absorbable sutures in conventional buffer media (9), the most common surface morphological characteristic upon hydrolytic degradation of suture fibers is the formation of circumferential or/and longitudinal surface cracks that are consistent with the anisotropic characteristic of fibers. It is not fully understood at this stage how superoxide ion induced degradation could lead to such unusual surface morphology on Monocryl and Maxon sutures.

The biocompatibility of degradation products is usually not a problem because all existing absorbable sutures are made from the well-known biocompatible glycolide, lactide, and their derivatives. However, biocompatibility of degradation products also depends on the rate of their accumulation in the surrounding tissues. This implies that the ability of the surrounding tissues to actively remove and metabolize degradation products is essential. Such a metabolism depends on the extent of blood circulation in the tissue. A well-vascularized tissue could remove degradation products as fast as they are released from an absorbable suture and subsequently metabolized, which could minimize tissue reactions to degradation products.

Because of their ability to release degradation products, absorbable sutures have recently been studied as a vehicle to deliver a variety of biochemicals such as growth factors to facilitate wound healing or antibiotics to combat wound infection. This new approach would increase the value of absorbable sutures and extend their function beyond the traditional role of wound closure. A typical example is Vicryl Plus, which has an antimicrobial agent coating (10,11).

Biodegradation properties are usually examined *in vitro* or/and *in vivo*. In the *in vitro* environment, the most commonly used medium is physiological saline phosphate buffer of pH 7.44 at 37°C. However, other buffers such as Tris or other body fluids such as urine, bile, and synovial fluids have been used. Occasionally, microorganisms were deliberately incorporated into these media to examine the effect of microorganisms on biodegradation properties of absorbable sutures. In the *in vivo* environment, unstressed absorbable sutures are normally implanted in rat gluteal muscle for predetermined periods of implantation. However, the use of unstressed sutures and gluteal muscle site may not represent the real clinical environment that absorbable sutures normally experience (28). The sutures retrieved at various periods of immersion or implantation are then subject to evaluation of their mechanical and physical properties to assess their changes with time. The degree of absorption *in vivo* is evaluated by the change

in suture cross-sectional area, while the level of tissue reaction is assessed by either the histological method and/or enzyme histochemistry.

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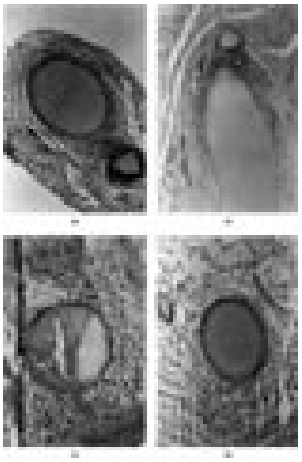

Fig. 1. Light histologic photomicrographs of tissue adjacent to PDS and Maxon sutures with 3 and 5 days post-implantation in a variety of tissues of New Zealand White rabbit (X130). (**a**): PDS in peritoneum; (**b**): PDS in fascia; (**c**): Maxon in peritoneum; (**d**): Maxon in fascia. From Ref. 29.

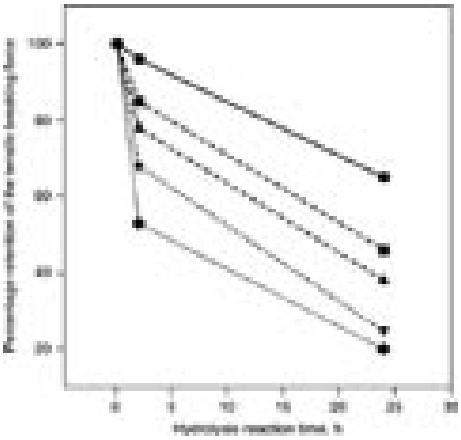

Fig. 2. The percentage of retention of tensile breaking force of five 2/0 synthetic absorbable sutures upon 0.005 *M* superoxide ion-induced hydrolytic degradation at 25°C. ● PDSII; ■ Dexon; ◆ Monocryl; ∫ Vicryl; | Maxon. From Ref. 35.

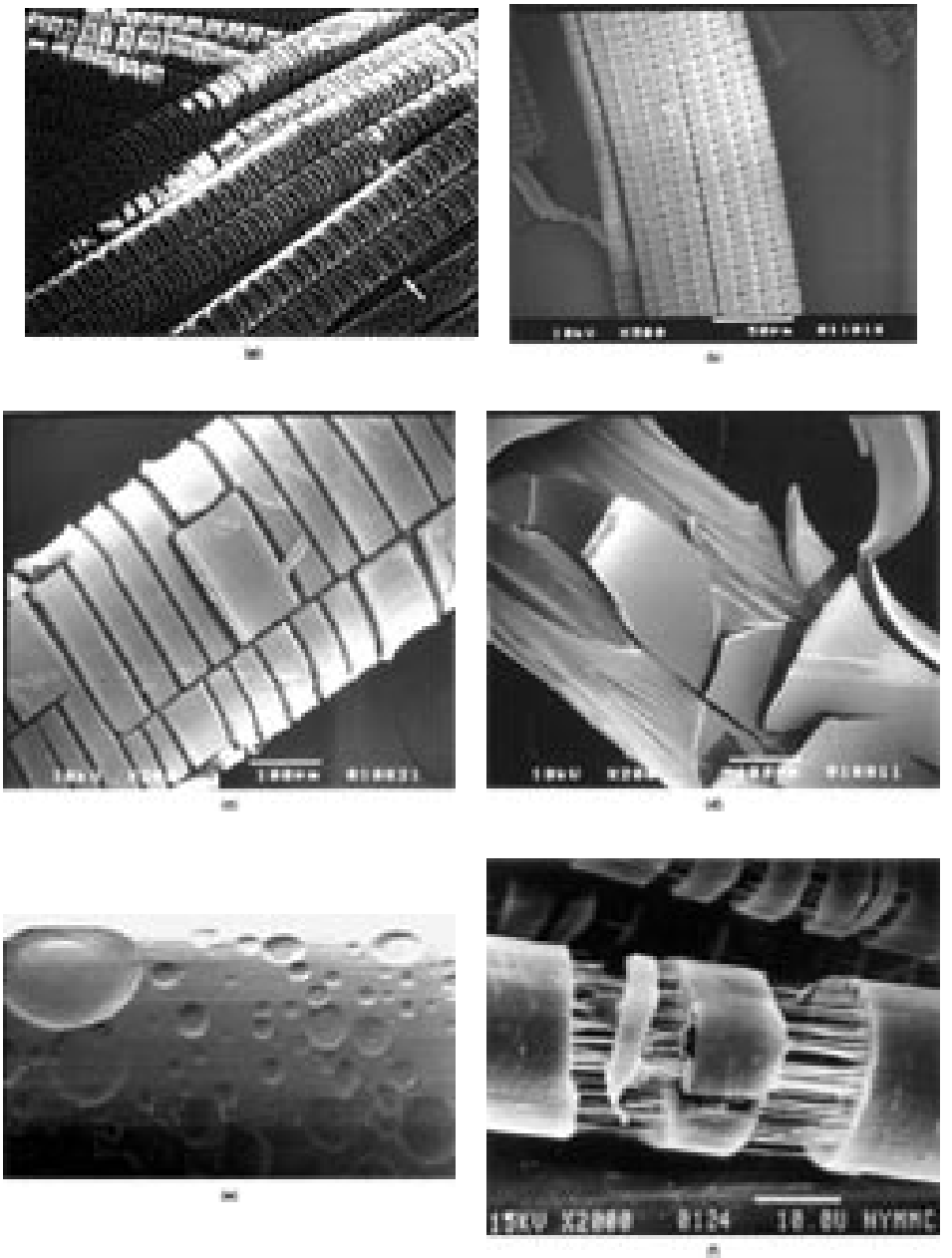

Fig. 3. Scanning electron images of degradation of some commercial absorbable sutures. (**a**) 2/0 Dexon after 20 Mrad γ irradiation at room temperature and 40 days *in vitro* buffer at 37°C; (**b**) 2/0 Dexon after 10 Mrad γ irradiation at 55°C and 7 days *in vitro* buffer at 37°; (**c**) 2/0 Maxon after 10 Mrad γ irradiation at 55°C and 42 days *in vitro* buffer at 37°C; (**d**) 2/0 Maxon after 2 Mrad γ irradiation at 55°C and 42 days *in vitro* buffer at 37°C; (**e**) 2/0 Monocryl upon 0.005 *M* superoxide ion-induced hydrolytic degradation at 25°C; (**f**) Polyglycolide fibers as the component of woven vascular grafts upon *in vitro* hydrolytic degradation in buffer of pH 7.4 at 37°C.

Table 1. **List of Commercial Sutures, Trade Names, and Manufacturers**

Generic name	Trade name	Physical configuration	Surface treatment	Manufacturer
Natural Absorbable Sutures				
catgut			plain and chromic	Surgical Specialties Corp.
catgut			plain and chromic	Dynek Sutures
catgut		Multifilament	monofilament finish, plain and chromic	SURU
collagen – bovine	Surgical Gut		plain, chromic, and mild chromic	USS/DG
collagen – bovine, ovine	Surgical Gut		plain and chromic	Ethicon
Synthetic Absorbable Sutures				
glycolic acid and trimethylene carbonate copolymer	Maxon	monofilament	clear or dyed green	USS/DG
glycolic acid and trimethylene carbonate copolymer	Maxon CV	monofilament	clear or dyed green	USS/DG
glycolic acid homopolymer	Dexon II	braided	dyed green or bicolored or undyed; coated with polycaprolate	USS/DG
glycolic acid homopolymer	Dexon S	braided	dyed green or undyed	USS/DG
glyconate	Monosyn	mid-term braided	dyed violet or undyed	B. Braun Melsungen AG
lactomer	Polysorb	braided	dyed violet or undyed	United States Surgical
poliglecaprone 25	Monocryl	monofilament	dyed or undyed	Ethicon
polydioxanone	PDS II	monofilament	dyed or undyed	Ethicon
polyester – glycolide (60%), dioxanone (14%), trimethylene (26%)	Biosyn	monofilament	dyed violet or undyed	USS/DG
polyglactin 910 at 90% glycolide and 10% lactide	Coated Vicryl	braided	dyed or undyed	Ethicon
polyglactin 910 at 90% glycolide and 10% lactide but irradiation-treated	Vicryl Rapide	braided	undyed	Ethicon
polyglactin 910 at 90% glycolide and 10% lactide	Vicryl Plus	braided	dyed violet or undyed; coated with glycolide and lactide (polyglactin 370) and calcium stearate	Ethicon
poly(glycolide-*co*-L-lactide) at 95% L-lactide and 5% glycolide composition ratio	Panacryl	braided	undyed; coated by a copolymer of ε-caprolactone and glycolide at 90 to 10 ratio	Ethicon

Table 1. (*Continued*)

Generic name	Trade name	Physical configuration	Surface treatment	Manufacturer
polyglycolic acid	PolySyn FA	braided	undyed	Surgical Specialties Corp.
polyglycolic acid	Surucryl	braided	coated with polycaprolactone calcium stearate	SURU
polyglycolic acid – low molecular weight	Safil Quick	mid-term braided	coated	B. Braun Melsungen AG
polyglycolic acid – pure	Safil Green	mid-term braided	coated; dyed green	B. Braun Melsungen AG
polyglycolic acid – pure	Safil Violet	short-term braided	coated; dyed violet or undyed	B. Braun Melsungen AG
polyglytone 6211 synthetic polyester	Caprosyn	monofilament	undyed	United States Surgical
poly-*p*-dioxanone	MonoPlus	long-tem monofilament	dyed violet	B. Braun Melsungen AG
	BioSorb	braided	dyed green	Alcon Labratories
	BioSorb Coated	braided	dyed green; coted with polycaprolate	Alcon Labratories
Nonabsorbable Sutures				
316L stainless steel	Surgical Stainless Steel	monofilament and multifilament		Ethicon
corrosion-resisted steel	Steelex	twisted or monofilament		B. Braun Melsungen AG
fibroin – natural	Sofsilk	braided	dyed black with logwood extract; coated with special wax	USS/DG
linen	Linatrix	twisted	natural white	B. Braun Melsungen AG
nylon	Nylene	monofilament	dyed blue	Dynek Sutures
nylon		monofilament	dyed black	Surgical Specialties Corp.
nylon-6 and -66	Ethilon		dyed green or black and Undyed clear	Ethicon
nylon-6 and -66	Monosof	monofilament	dyed black with logwood extract or undyed; silicone-coated	USS/DG
nylon-6 and -66	Dermalon	monofilament	dyed blue; silicone-coated	USS/DG
nylon-6 and -66	Surgilon	braided	dyed blue; silicone-coated	USS/DG
nylon		monofilament	dyed black	Alcon Labratories
polyethylene (terephthalate)	TI·RON	braided	dyed blue or undyed; silicone-coated or uncoated	USS/DG
poly(vinylidene fluoride) and polyvinylidene fluoride-*co*-hexafluoropropylene)	Pronova		dyed blue	Ethicon
polyamide (nylon)	Surulon	monofilament	dyed blue/black	SURU
polyamide 6	Dafilon	monofilament	dyed blue	B. Braun Melsungen AG
polyamide 66	Dafilon	monofilament	dyed black	B. Braun Melsungen AG

polyamide 6 66	Supramid	pseudomonofilament-core polyamide 6.6	cover- polyamide 6	B. Braun Melsungen AG
polybutester	Novafil	monofilament	dyed and undyed	USS/DG
polybutester	Vascufil	monofilament	dyed blue; coated	USS/DG
poly(butylene terephthalate) polyester	Miralene	monofilament	dyed blue	B. Braun Melsungen AG
polyester	Surgidac	braided	dyed greem; coated	USS/DG
polyester	Surupol	braided	dyed green/white, silicone-coated	SURU
polyester		braided or monofilament	dyed white or green	Alcon Labratories
polyester	Polyviolene	braided	dyed green or white	Surgical Specialties Corp.
polyether	Dyloc	monofilament	dyed blue	Dynek Sutures
poly(ethylene terephthalate) polyester	Dagrofil HRT	braided	uncoated; dyed green	B. Braun Melsungen AG
poly(ethylene terephthalate) polyester	Synthofil	braided	coated; dyed green or undyed	B. Braun Melsungen AG
poly(ethylene terephthalate) polyester	PremiCron	braided	silicone-coated; dyed green or white	B. Braun Melsungen AG
poly(ethylene terephthalate) polyester	Ethibond	braided	dyed green; coated with polybutilate or poly	Ethicon
polypropylene	Premilene	monofilament	dyed blue with coppet phtalocyanine	B. Braun Melsungen AG
polypropylene	Prolene		dyed blue	Ethicon
polypropylene	Surgipro	monofilament	dyed blue or undyed	United States Surgical
polypropylene	Surgipro II	monofilament	dyed blue or undyed	United States Surgical
polypropylene	Surulene	monofilament	dyed blue	SURU
polypropylene		monofilament	dyed blue	Alcon Labratories
polypropylene		monofilament	dyed blue	Surgical Specialties Corporations
poly(vinylidene fluoride)	Radene	monofilament	dyed blue	Dynek Sutures
poly(vinylidene fluoride)	Vilene	monofilament	dyed blue	Dynek Sutures
siliconized polyester	Polyflex	braided	dyed black	Dynek Sutures
siliconized polyester	Dyflex	braided	dyed green	Dynek Sutures
siliconized polyester	Teflex	braided	white	Dynek Sutures
silk	Surusil	braided	dyed black, treated with wax	SURU
silk		twisted or braided	dyed black or white	Alcon Labratories
silk		braided	dyed black	Surgical Specialties Corporations
silk – fibroin	Perma-Hand	natural waxes and gums removed	dyed black and undyed; special wax	Ethicon
silk – fibroin	Perma-Hand – virgin	sericin gum not removed		Ethicon
silk – natural	Silram	braided	coated: dyed black	B. Braun Melsungen AG

Table 1. (*Continued*)

Generic name	Trade name	Physical configuration	Surface treatment	Manufacturer
silk – natural	Virgin Silk	twisted	dyed methylene blue	B. Braun Melsungen AG
stainless steel	Flexon	multifilament	FEB polymer coating	USS/DG
stainless steel	Steel	monofilament		USS/DG
treated silk	Dysilk	braided	dyed black	Dynek Sutures
USP8/0 to 10/0 in various materials	Microflex			Dynek Sutures

Table 2. **Four Essential Properties of Suture Materials**

Physical/Mechanical	Handling	Biocompatibility	Biodegradation
USP vs EP size (diameter)	pliability	inflammatory reaction	tensile breaking strength and mass loss profiles
mono vs multifilament	memory		
tensile breaking strength and elongation	knot tie-down knot slippage	propensity toward wound infection, calculi formation, thrombi formation, carcinogenicity, allergy	
modulus of elasticity	tissue drag		biocompatibility of degradation products
stiffness			
stress relaxation and creep			
capillarity			
swelling			
coefficient of friction			

Table 3. **Absorption Delay of Commercial Synthetic Absorbable Sutures**

Suture materials	Time to complete loss of tensile strength, days	Time to complete mass absorption, days	Absorption delay, days	Useful Lifetime[a], (%)
Dexon	28	50–140	22–112	20–56
Vicryl[b]	28	90–120	23–62	23–31
Panacryl	365[c]	548–916	183–551	39–66[c]
PDS	63	180–240	117–170	26–35
Maxon	56	210	155	27
Monocryl	21	90–119	69–98	18–23

[a]The ratio of (the time to complete loss of tensile strength) to (the time to complete mass absorption) × 100. The higher the percentage is, the better absorbable the suture.
[b]Does not include Vicryl Rapide, which is completely absorbed within 35 days in a living tissue.
[c]Based on 20% original strength remaining rather than 100% strength loss.