Acta of Bioengineering and Biomechanics
Vol. 27, No. 3, 2025

Original paper
DOI: 10.37190/abb/214162

Evaluation of the materials properties used for the production of surgical threads during the simulation of use in eye microsurgery

MATEUSZ SOBIERAJ[1], MACIEJ ŁUCZAK[1], NATALIA KIEŁPIŃSKA[2], ARTUR KOŚCIUSZKO[2]

[1] Bydgoszcz University of Science and Technology, Bydgoszcz, Poland.
[2] Bydgoszcz University of Science and Technology, Faculty of Mechanical Engineering, Department of Manufacturing Techniques, Bydgoszcz, Poland.

Purpose: The aim of this study was to evaluate absorbable and non-absorbable surgical sutures exposed to an environment with a chemical composition similar to that of ocular body fluids. *Methods*: The evaluation was based on the results of tests of the mass, diameter and mechanical properties of samples immersed in physiological saline solution (BSS) at different time intervals. *Results*: Based on the conducted research, it was found that multifilament threads made of PGCL dissolve the fastest under these conditions, while PDS monofilament threads dissolve the longest. In the first case, the last measurements could be taken after 14 days of immersion, while in the second case, the monofilament was not completely dissolved even after 80 days. *Conclusions*: Despite numerous publications in this area, available from various sources, it is very difficult to compare the obtained results to those of other authors. This is due to the fact that studies conducted on threads made of different materials and diameters, as well as in different environments, can have a significant impact on the resorption process. This justifies the need for this type of research.

Key words: *surgical threads, eye microsurgery, mechanical properties, resorption*

1. Introduction

Surgery is one of the oldest and most important fields of medicine. The development of surgery has always been closely related to the improvement of surgical techniques, the development of medical devices and the refinement of materials commonly used in medicine. Sutures are essential for surgical procedures, enabling precise tissue suturing and supporting wound healing [6], [8]. Currently, many types of sutures with varying properties, tailored to surgical needs, are available on the market. The choice of suture depends on the type of surgery, the properties of the tissues being sutured and the required support time. The decision on which thread should be used in a given procedure is made by the surgeon [10], [13].

One of the main criteria for classifying sutures from clinical perspective is their absorbability. They are divided into absorbable and non-absorbable sutures [9], [18], [42]. Absorbable ones are also called resorbable sutures. The materials used for their production are of natural origin, e.g., collagen [42], surgical catgut [10] or synthetic origin, including polydioxanone, polyglactin 910, and poliglecaprone 25 sutures [33]. Absorbable sutures degrade, losing approximately 50% of their tensile strength after 60 days. Non-absorbable sutures, on the other hand, maintain their tensile strength beyond 60 days [8], [9]. Non-absorbable and synthetic sutures include polypropylene and nylon ones [33], [42], and natural threads include silk and surgical cotton [8], [9].

Surgical sutures can also be classified based on the number of fibers. Monofilament and multifilament sutures are distinguished. Monofilament sutures are made

* Corresponding author: Natalia Kiełpińska, Bydgoszcz University of Science and Technology, Faculty of Mechanical Engineering, Department of Manufacturing Techniques, Bydgoszcz, Poland, e-mail: natalia.kielpinska@pbs.edu.pl
Received: September 19th, 2025
Accepted for publication: November 12th, 2025

from a single fiber. They are characterized by an uniform and smooth structure and resistance to the settlement and penetration of pathogenic microorganisms [6], [10], [25]. Furthermore, they are characterized by lower elasticity and resistance to knot formation. Multifilament sutures consist of more than one fiber. The fibers within the structure can be twisted or braided. Multifilament threads are characterized by higher elasticity, flexibility, and tensile strength than monofilamnets ones [9], [25]. The surface of multifilament sutures is rough, which causes greater friction against the tissue and greater susceptibility to infection compared to monofilament sutures [9], [25].

The selection of suture materials is crucial due to their direct contact with the human body. Materials must meet a number of criteria, including biocompatibility, physicochemical properties, mechanical parameters and usability [42]. Requirements for surgical sutures include high and uniform tensile strength, enabling the use of lower diameters threads. Furthermore, the ability to maintain the initial tensile strength in the body is crucial, enabling wound support throughout the critical healing period, followed by rapid absorption. High elasticity and flexibility, enabling bending and knot tying, sterility and a lack of biological activity, including carcinogenic activity, are also important [11], [18], [24]. For absorbable surgical sutures, manufacturers provide the total absorption time and effective tissue support period. For example, according to the manufacturer [12], PGLA threads have a total absorption time of 56–70 days, while the effective tissue support time is up to 35 days. For the PGA material [19], the thread may degrade within 60–90 days, and the effective tissue support time after 3 weeks is 35% of the initial value.

In recent years, an important area of development in surgical sutures has been the search for new materials or the modification of existing ones [3], [40], taking into account their area of application, biocompatibility, and mechanical properties. Guambo et al. [15] studied cellulose coconut (*Cocos nucifera*) and sisal (*Agave sisalana*) fibers. Kampeerapappun et al. [20] extracted cellulose fibers from cylindrical snakegrass (*Dracaena angolensis*) and then tested them to assess their potential use as a non-absorbable biomaterial for surgical sutures. A material based on the silk of the spider Argiope bruennichi was studied by Turan et al. [39].

Drug-coated surgical threads are also being produced and tested to support the wound healing process and release anti-inflammatory, or analgesic drugs at the suture site [8], [16], [17], [38]. Sriyai et al. [36] developed an absorbable monofilament suture made of poly(*l*-lactide-co-ε-caprolactone) coated with an antibacterial coating. Yang et al. [41] prepared a surgical thread consisting of a PLA filament core coated with nanofibrous membranes composed of polyglycolic acid (PGA), polycaprolactone (PCL) and an antibacterial drug.

Research is also being conducted on surgical sutures in conditions simulating their application environment [7], [30]. Ojastha and Jeevitha evaluated the tensile strength of polyglactin sutures after immersion in herbal mouthwashes [31].

The degradation of surgical sutures made of PDS, PGLA and PGCL was studied in bile, pancreatic juice, in sterile form, and contaminated with bacteria. Both sutures containing and without an antibacterial triclosan coating were tested. Physiological saline was used as a reference. Studies by Merkel et al. [28] demonstrated that sutures made of PDS were characterized by higher resistance to degradation, while maintaining their mechanical properties for a longer period of time compared to PGLA and PGCL sutures. Furthermore, they observed that the antibacterial coating on the sutures delayed the degradation process.

A study by Akti et al. [2] examined the effect of artificial saliva and chlorhexidine mouthwash on the tensile strength of absorbable PGLA sutures and non-absorbable silk sutures. The authors' study demonstrated that the use of this mouthwash can significantly reduce suture resistance after 14 days.

The effect of artificial saliva versus saline on the rate of suture degradation in oral and pharyngeal surgery was analyzed. The study was conducted on absorbable chromic sutures, poliglecaprone 25, and poliglactin 910. Based on the study, Briddell et al. [5], indicated that all suture types tested degraded more rapidly in artificial saliva compared to saline. Differences in suture degradation were also observed depending on the material.

In ophthalmic procedures and surgeries, surgical threads are used for various types of procedures, primarily for precise tissue suturing during cataract surgery (for closing corneal and scleral incisions), corneal transplants (keratoplasty), in anti-glaucoma procedures such as trabeculectomy, as well as in eyelid surgery and suturing of lesions (corneal, scleral, conjunctival wounds).

In ocular muscle surgery (e.g., strabismus treatment), threads are used to adjust the muscle position to ensure precise eyeball positioning. During vitrectomy, threads may be needed to close working ports after trocar removal, especially if the ports are leaky and there is a risk of fluid leakage or ocular hypotension. In these procedures, it is essential to select very thin threads (11/0–7/0), which minimize trauma, scarring and gently approximate the tissues. Materials used for threads include nylon and polyglactin 910 [27], polypropylene, PLLA/PEG, PLGA, PCL, poly(tetrafluoroethylene), and ultra-high molecular weight polyethylene [35].

Matalia et al. [27] conducted a study to compare monofilament nylon threads and 10-0 polyglactin 910 threads in terms of effectiveness as well as the frequency and types of suture-related complications in cataract surgery in children.

Agarwal et al. [1] also conducted research to evaluate and compare complications associated with the use of 10-0 nylon sutures and 10-0 polyglactin sutures in pediatric cataract surgery. Kashiwabuchi et al. [22] developed and tested an absorbable surgical suture made of poly(L-lactide), polyethylene glycol, which released antibiotics for use in ophthalmic procedures.

The aim of this study was to evaluate the properties of materials used to produce surgical sutures during conditioning in Balanced Salt Solution (BSS) in the context of their potential use in ophthalmic surgery. The durability of the sutures was assessed during exposure to environmental conditions similar to those of the ocular fluids. The study focused solely on material aspects; therefore, no analysis of medical issues was undertaken.

2. Materials and methods

2.1. Materials

The study used surgical sutures made of six different materials and from different manufacturers, which also differed in structure (monofilament and multifilament) and resorption in the human body (absorbable and non-absorbable). A common feature of all surgical sutures was their USP 1 size, meaning the suture diameter should be between 0.400 mm and 0.499 mm. The general characteristics of the sutures used in the study are presented in Table 1. The sutures' properties measured before the conditioning process are presented in Table 2. The diameters of some sutures do not precisely fall within this size range, which may be due to the measurement methodology used. In ophthalmology, sutures with larger USP sizes and smaller diameters are used. However, material testing of these structures requires highly precise measuring devices. Therefore, it was decided to use the largest diameter sutures for the measurements. This was done to observe phenomena occurring during immersion.

2.2. Samples preparation

To simulate operating conditions, surgical sutures were placed in containers with Balances Salt Solution (BSS) (Aqueo Premium™, BVI), a saline solution specifically developed for ophthalmic applications, including eye surgery. The chemical composition of the BSS solution (Table 3) is physiologically matched to the aqueous humor of the eye and contains more electrolytes than standard 0.9% NaCl saline. The solution was neutral, with a pH of 7–7.5. Samples were conditioned at 37 °C ± 1 °C. The conditioning temperature was close to human body temperature, and the conditioned time was 1, 7, 14, and 30 days, which correspond to the

Table 1. Types of surgical sutures used in the studies

Type of material	Signtaure	Resorption	Structure	Producer
Polyamide	PA	non-absorbable	monofilment	Jinhuan Medical Products
Polyethylene terephthalate	PET	non-absorbable	multifilament	YAVO
Polydioxanone (antibacterial)	PDS	absorbable	monofilment	Ethicon
Polyglycolic acid	PGA	absorbable	multifilament	Jinhuan Medical Products
Poliglecaprone 25 (Monocryl)	PGCL	absorbable	monofilment	Ethicon
Polyglactin 910 (Vicryl)	PGLA	absorbable	multifilament	Ethicon

Table 2. Thread properties before the conditioning process

Material	Sample mass [mg]	Elastic modulus [MPa]	Tensile strength [MPa]	Elongation at break [%]
PA	16.39	1930	192	18.9
PET	24.67	5480	301	10.8
PDS	35.85	982	281	31.0
PGA	24.06	7300	428	14.9
PGCL	34.52	618	301	23.5
PGLA	27.27	7660	332	10.2

tissue retention times declared by the manufacturers, and at 80 days to understand the properties of the materials after long-term exposure. Additionally, tests were carried out on reference samples that were not conditioned in the BSS solution (conditioning time: 0 days).

Table 3. Chemical composition of BSS physiological solution in 1000 ml

Chemical component	Mass [g]
Sodium chloride	6.40
Potassium chloride	0.75
Calcium chloride dihydrate	0.48
Magnesium chloride hexahydrate	0.30
Sodium acetate trihydrate	3.90
Sodium citrate dihydrate	1.70

2.3. Samples mass measurements

The mass of the conditioned samples was measured using a Sartorius CPA225D-0CE analytical balance with an accuracy of 0.01 mg. Before measurement, the thread sections were dried from the BSS solution using absorbent paper. The tests were performed on 10 samples of each type at 23 °C ± 2 °C. The length of the tested sections was 90 mm.

2.4. Optical microscopy

The study of the structure evaluation and diameter measurement of surgical sutures was performed on a Keyence VHC-7000 digital optical microscope (Osaka, Japan) equipped with a VH-Z100R objective lens at a magnification of ×200. Measurements were performed for 10 samples at a temperature of 23 °C ± 2 °C in depth-of-field mode.

2.5. Testing of mechanical properties

Mechanical properties of conditioned surgical sutures were tested during static tensile testing using a Z030 universal tensile testing machine from ZwickRoell Roell (Ulm, Germany). The tensile rate was 1 mm/min to determine the elastic modulus, and then 50 mm/min until the samples broke. The measuring section was 50 mm long. Measurements were performed for 10 samples of each type at a temperature of 23 °C ± 2 °C. The total length of the tested samples was 90 mm. To determine the tension, the thread cross-sectional area values calculated from microscopic measurements from each day of measurement were used. In most publications in this field, the authors use the maximum breaking force parameter and also conduct tests on samples with surgical knots [2], [5], [42]. In this study, mechanical properties were determined during static tensile testing in the form of stress expressed in MPa.

3. Results

3.1. Samples mass

Graphs of the weight change of surgical suture samples during immersion in BSS (physiological saline solution) are presented in Fig. 1. In the case of non-absorbable PA sutures, the weight during conditioning was stable, despite the fact that this polymer is characterized by a relatively high water absorption (approximately 10%). An interesting result was also recorded for non-absorbable PET sutures, whose weight increased by less than 10 wt.% over 80 days of conditioning.

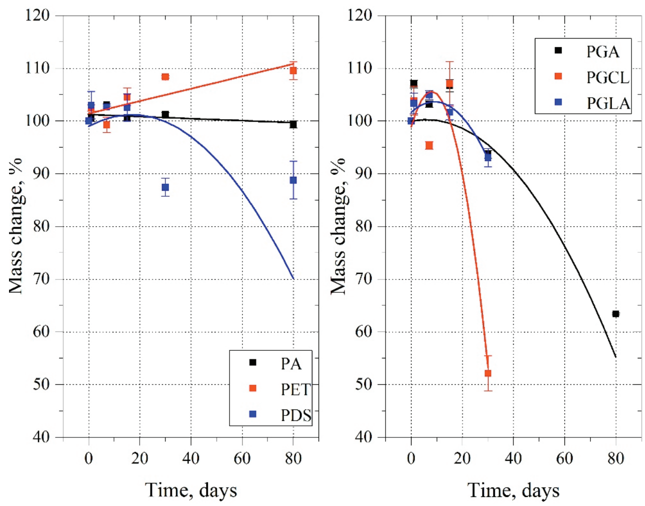

Fig. 1. Change in the mass of surgical threads during conditioning

For absorbable sutures, the expected weight loss over time is preceded by the opposite phenomenon. By day 14, all sutures exhibited a higher weight than the initial weight. This is the result of water absorption and adsorption in the case of multifilament sutures. In the case of PGA and PGCL, the weight increase was up to 7 wt.%. After 14 days, a decrease in the weight of all absorbable sutures was observed, indicating the initiation of the dissolution process. Complete dissolution of PDS and PGA sutures took longer than the time specified in the study methodology. After 80 days, the weight loss of these materials was 12 wt.% (PDS) and 37 wt.%

(PGA), respectively. The dissolution rate of PGLA and PGCL sutures is significantly higher. On day 30 of conditioning, the sample weights were 93 wt.% (PGLA) and 52 wt.% (PGCL), respectively. After 80 days, the samples were almost completely dissolved, and their mass measurement was impossible.

3.2. Optical microscopy

Absorption and dissolution of surgical sutures during conditioning resulted in changes in their diameter in most cases (Fig. 2). The exceptions are PA and PET sutures, whose diameter does not change substantially over time because these fibers are non-absorbable and these phenomena are less noticeable in these cases. The stable dimensions of the PET sutures also confirm that the mass increase described above resulted from liquid adsorption by the multifilament, not absorption. In the case of resorbable sutures, dissolution was preceded by water absorption, as indicated by the increase in sample mass. This phenomenon was accompanied by thread swelling, confirmed by an increase in their diameter during the initial conditioning period. In the case of PDS, the diameter increase was approximately 2.5%, however, as a result of water absorption, the diameter of the PGA and PGLA samples increased by over 20%. Interestingly, the diameter of the monofilament suture made of PGCL did not increase despite the increase in mass. This may be due to the dissolution rate, which was the fastest for this material. Microscopic photographs of selected resorbable sutures before and after 14 days of conditioning (PDS, PGLA, PA) are presented in Fig. 3. After 14 days, the monofilament PDS suture exhibited numerous microcracks on the surface of the material, indicating progressive degradation of the material. The multifilament PGLA suture showed no significant changes due to multi-day immersion in the BSS solution.

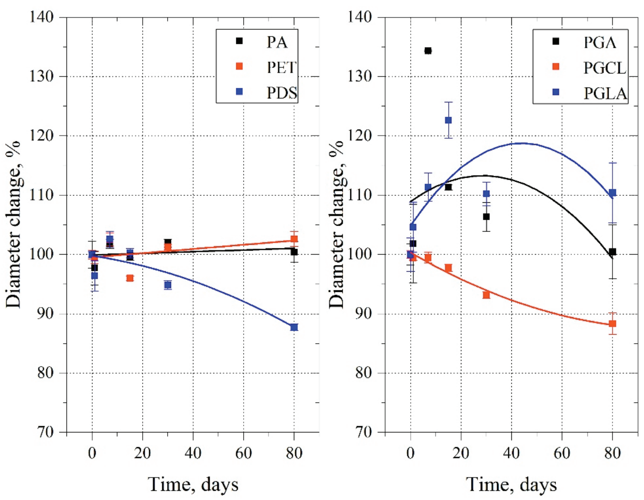

Fig. 2. Change in diameter of surgical threads during conditioning

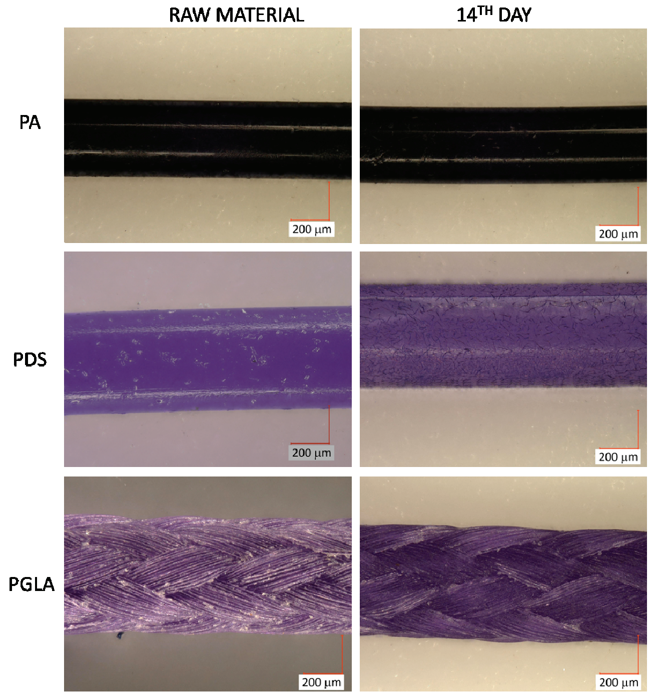

Fig. 3. Photographs of PA, PDS and PGLA surgical sutures before and after 14 days conditioning

3.2. Mechanical properties

The mechanical properties of surgical sutures change during conditioning, regardless of whether the suture is soluble or insoluble. In the case of soluble threads, tests after 80 days of conditioning were only performed for the PDS material. In the remaining cases, the samples were either completely dissolved or broke when clamped between the grips of the tensile testing machine.

The elastic modulus (Fig. 4) of PA and PET threads decreases by approximately 40–50% within the first 24 hours and remains unchanged during further immersion at 37 °C. In contrast, the PDS monofilament maintains its initial properties until day 30 of conditioning. Then, on day 80, the elastic modulus decreases to 20% of the initial value.

Changes in stiffness during conditioning for the remaining materials are similar and occur significantly faster than for PDS. After just 30 days of immersion, 20% of the initial elastic modulus was recorded for PGA and PGLA. The mechanical properties of PGCL sutures could only be determined up to day 15.

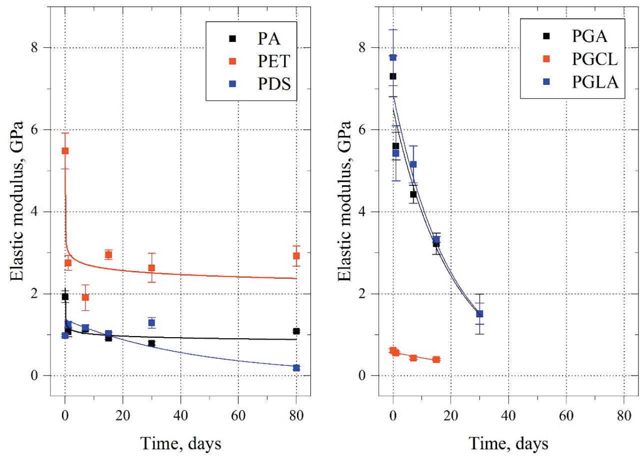

Fig. 4. Change in the elastic modulus of surgical threads during conditioning

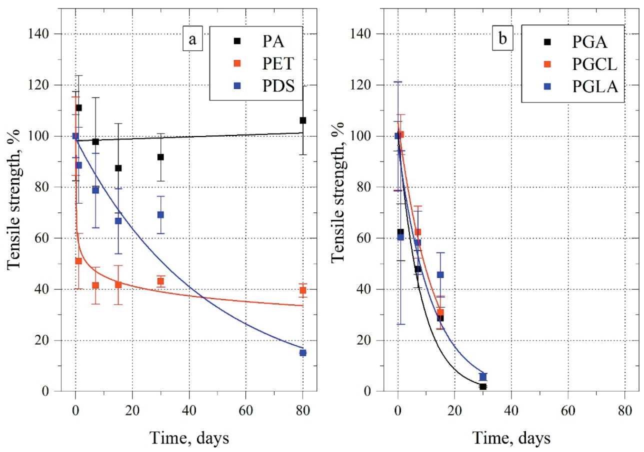

Fig. 5. Change in the tensile strength of surgical sutures during conditioning

The tensile strength (R_m) of the non-absorbed PA thread remains constant throughout the immersion period (Fig. 5). The R_m of the PET multifilament decreased by approximately 50% after the first 24 hours, reaching 40% of the initial value after 80 days.

The rate of decline in the tensile strength of PDS threads during the initial conditioning phase is not as rapid as that of PET. However, the R_m value of this material does not stabilize and after 80 days of immersion reaches a value below 20% of the initial value. The rate of change in the tensile strength of the remaining absorbable threads is greater than that of PDS. After 30 days, PGA and PGLA threads exhibit R_m values below 10% of the initial value, while for PGCL, measurements could not be obtained after this time, as previously described. The absolute values of the diameter and maximum load of surgical sutures after conditioning are collected in Table 4.

The strain at break of the non-absorbed threads remained constant throughout the conditioning period (Fig. 6). The strain of the absorbable threads decreased with immersion time, reaching values of 5% and 15% of the initial value for PGA and PGLA, respectively. However, 15% of the initial strain at break for PDS was recorded only on day 80 of conditioning.

In Figure 7, sample tensile stress curves for threads before and after 14 days of conditioning are shown. The absence of a yield point, coupled with high maximum stress values, indicates that these products were subjected to a tensile process during production, resulting in macromolecular orientation. The maximum stresses of conditioned threads are always lower than those of raw threads, even in the case of insoluble threads. Note the

Table 4. Absolute values of the maximum load and diameter of surgical threads

Material	Parameter*	Conditioning time [days]					
		0	1	7	14	30	80
PA	F_{max} [N]	23.75	25.2	24.1	20.6	22.6	25.4
	D [mm]	0.397	0.388	0.404	0.395	0.405	0.399
PET	F_{max} [N]	72.00	23.2	20.1	17.7	20.3	19.2
	D [mm]	0.552	0.549	0.566	0.530	0.559	0.566
PDS	F_{max} [N]	65.76	37.0	37.3	30.1	28.0	5.2
	D [mm]	0.546	0.526	0.560	0.547	0.518	0.479
PGA	F_{max} [N]	66.83	19.4	25.9	10.7	0.60	-
	D [mm]	0.446	0.454	0.599	0.497	0.474	0.448
PGCL	F_{max} [N]	68.65	43.6	27.0	13.0	-	-
	D [mm]	0.539	0.536	0.536	0.527	0.502	0.476
PGLA	F_{max} [N]	52.31	20.0	21.8	20.8	2.0	-
	D [mm]	0.448	0.469	0.499	0.550	0.494	0.495

* F_{max} – maximum load, D – diameter.

significantly lower tensile strength of the conditioned PET multifilament, while there is no significant change in the strain at break.

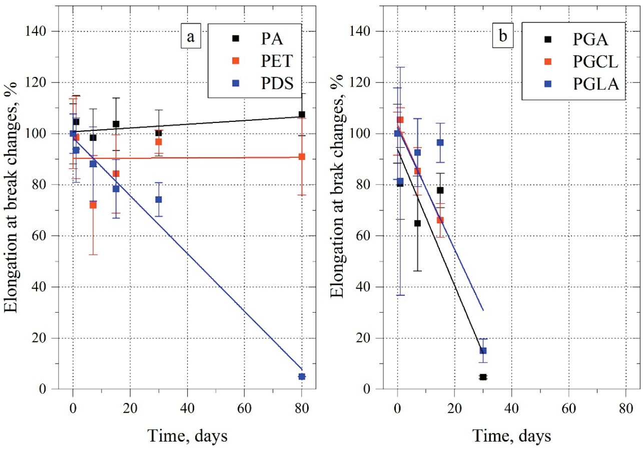

Fig. 6. Change in the elongation at break of surgical sutures during conditioning

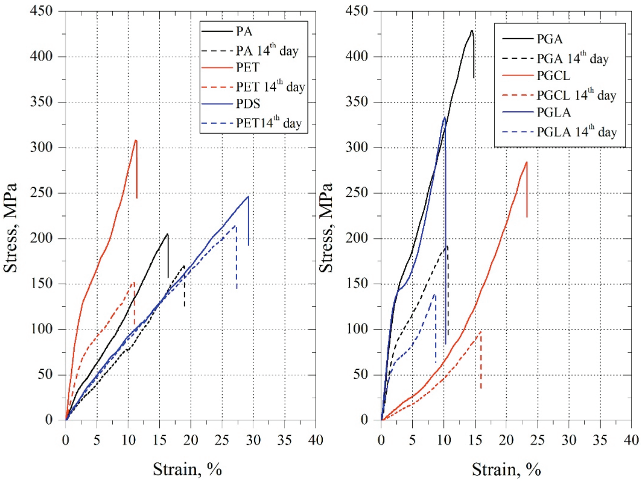

Fig. 7. Examples of surgical suture stretching curves

4. Discussion

The surprisingly high water absorption of PET threads described in the article is the result of liquid penetration between the individual filaments of the thread, which, unlike PA, had a multifilament structure. This is confirmed by the maximum water absorption value of this material, which is less than 1 wt.% [23].

The swelling of absorbable sutures in the initial conditioning phase was also observed by Barbosa [4]. However, in this case, these were chitosan and catgut sutures immersed in PBS (phosphate buffered saline) solution. An increase in mass at the beginning of the conditioning process was also observed for the PDS II material in the work of Ooi [32]. However, after 80 days, a 3% mass loss was observed. In this case, a USP 2-0 size suture was tested in potassium phosphate buffered saline. It can therefore be concluded that many factors influence suture degradation. This is indicated by the recorded 10% mass loss after 80 days of conditioning described in this work.

In another study [3], chitosan-coated PLA fibers conditioned in saline lost 50% of their weight after just 15 days of testing. However, in this case, biodegradation occurred at 70 °C. Furthermore, the degradation rate of absorbable sutures is influenced not only by temperature but also by the pH of the solution, as proved by Niculescu [30].

The rate of mass loss during hydrolytic degradation of PGA material conditioned in a buffer solution for 24 days at 37 °C is greater for smaller sample sizes, as demonstrated by Shawe in his study [34]. The study was conducted on 0.06 mm, 0.09 mm, 0.13 mm, and 0.3 mm thick tapes. The researcher indicated that the rate of mass loss depends on the surface-to-volume ratio. Therefore, it can be assumed that placing smaller diameter surgical sutures in the solution will contribute to a shorter dissolution time due to their lower surface-to-volume ratio compared to larger diameter sutures, and thus a shorter diffusion path.

Changes in the mechanical properties of surgical sutures, both soluble and insoluble, during conditioning have also been observed by other authors [2], [42]. A stable value of the PDS elastic modulus in the initial phase of conditioning was also observed by other researchers for threads without an antibacterial coating [26], [41]. However, it is difficult to find in other publications the results of mechanical tests after 80 days of conditioning, and after 30 days of immersion in physiological saline solution, the elastic modulus decreased significantly. Interestingly, Müller [29] observed an increase in stiffness of PDS sutures starting from day 42 of the study. An increase in stiffness was also noted between days 14 and 42 of conditioning in Ringer solution for PGLA sutures.

The significant reduction in the tensile strength of PET multifilament is surprising, especially since in the publication [21], no statistically significant changes in the strength of the polyester threads were observed. In the same study, a similar relationship was observed for PA, despite the fact that conditioning was carried out in Ringer's solution at 22 °C. In the study [42], a different relationship was observed for PA threads. After 14 days of conditioning in Ringer's solution, a 20% increase in the force required to break the thread was noted.

The lowest rate of change in strength over time was observed for PDS with an antibacterial coating. This trend was also observed by Gierek [14]. Furthermore, after 28 days of conditioning in physiological saline, they observed a decrease in R_m for PGLA threads to 21.5% of the initial value, and for PGCL threads, they achieved approximately 40% of the initial R_m value on day 14.

In the work of Szabelski [37], similarly to the studies described here, a decreasing tendency of the strain at break was observed for the polyglycolic acid suture during conditioning in Ringer's solution. However, in the case of the PDS suture without antibacterial coating, the researchers observed a significant increase in the mean strain at break during conditioning.

5. Conclusions

Surgical sutures vary in their materials, size, and structure. They exhibit different behaviors depending on the chemical composition, temperature, and pH of the environment in which they are used. Furthermore, different researchers use different research methodologies to determine their mechanical properties. Therefore, it is extremely difficult to compare the results presented in different scientific publications.

Based on the conducted studies, it was found that the multifilament PGCL thread exhibited the highest resorption rate in BSS solution at 37 °C, while the PDS monofilament thread exhibited the longest rate. In the first phase of the resorption process, all tested threads absorbed water, which in most cases was accompanied by swelling. Surprisingly, the tensile strength of the PET threads decreased significantly after just 24 hours of conditioning. The resorbable PGA and PDS threads did not dissolve until day 80 of the study, similarly to the nonresorbable PA and PET threads. However, in the case of PGA, the degree of degradation was so advanced that mass loss could be assessed, but mechanical property measurements were not possible.

In eye microsurgery, monofilament sutures are primarily used due to their smoother surface and reduced risk of microbial colonization. During cataract surgery in adults, sutures with stable mass and minimal changes in diameter and strength are preferd for emergency closure of leaky incisions. Among the materials tested, polyamide meets these criteria, demonstrating the greatest stability of its properties. For conjunctival closure, when a longer-lasting, absorbable support is required, the best solution is PDS, which maintains its mechanical properties the longest among the absorbable sutures tested.

References

[1] AGARWAL P., MAAN V., SUTAR S., CHAUHAN L., *Suture Selection for Incision Closure in Pediatric Cataract Surgery: A Dilemma for Pediatric Ophthalmologists*, J. Pediatr. Ophthalmol. Strabismus, 2022, 59 (4), 243–247, DOI: 10.3928/01913913-20211111-02.

[2] AKTI A., CENGIZ Z.O., GÜRSES G., SERIN H., *Investigation of Absorbable and Non-Absorbable Multifilament Suture Materials in Terms of Strength Changes Using Chlorhexidine Mouthwash and Thermal Cycling: An In Vitro Study*, Materials, 2024, 17 (15), 3862, DOI: 10.3390/ma17153862.

[3] ALHULAYBI Z.A., *Fabrication and Characterization of Poly(lac tic acid)-Based Biopolymer for Surgical Sutures*, Chem. Engineering, 2023, 7 (5), 98, DOI: 10.3390/chemengineering7050098.

[4] BARBOSA MCDS., DA SILVA H.N., LOPES DDS. et al., *Biodegradable Chitosan Sutures Enhanced with N-Acetyl-D-Glucosamine: Comparative Study with Catgut Sutures*, Mat. Res., 2024, 27, DOI: 10.1590/1980-5373-mr-2024-0278.

[5] BRIDDELL J.W., RIEXINGER L.E., GRAHAM J., EBENSTEIN D.M., *Comparison of Artificial Saliva vs Saline Solution on Rate of Suture Degradation in Oropharyngeal Surgery*, JAMA Otolaryngol. Head Neck Surg., 2018, 144 (9), 824–830, DOI: 10.1001/jamaoto.2018.1441.

[6] BYRNE M., ALY A. *The Surgical Suture*, Aesthet. Surg. J., 2019, 39 (Suppl. 2), 67–72. DOI: 10.1093/asj/sjz036.

[7] CAWTHORNE D.P., CASTILLO T.E., SIVAKUMAR B.S., *Effect of commonly used surgical solutions on the tensile strength of absorbable sutures: an in-vitro study*, ANZ J. Surg., 2021, 91, 1451–1454, DOI: 10.1111/ans.16908.

[8] DENG X., QASIM M., ALI A., *Engineering and polymeric composition of drug-eluting suture: A review*, J. Biomed. Mater. Res. A., 2021, 109 (10), 2065–2081, DOI: 10.1002/jbm.a.37194.

[9] DENNIS C., SETHU S., NAYAK S., MOHAN L., MORSI Y.Y., MANIVASAGAM G., *Suture materials – Current and emerging trends*, J. Biomed. Mater. Res. A., 2016, 104 (6), 1544–1559, DOI: 10.1002/jbm.a.35683.

[10] DRESING K., SLONGO T., *Chirurgisches Nahtmaterial – Grundlagen*, Oper. Orthop. Traumatol., 2023, 35, 298–316, DOI: 10.1007/s00064-023-00812-y.

[11] Ethicon. (n.d.), *Wound closure manual*, Retrieved from https://anwresidency.com/simulation/guide/resources/Ethicon_Wound_Closure_manual.pdf (acces: 28.08.2024)

[12] Ethicon, *Katalog produktów szwy chirurgiczne, siatki przepuklinowe, hemostatyki*, 2014. Retrieved from https://anmar.tychy.pl/wp-content/uploads/2014/12/szwy_siatki_hemostatyki.pdf [Accessed: 30.07.2025].

[13] GIEREK M., KUŚNIERZ K., LAMPE P. et al. *Absorbable sutures in general surgery – review, available materials, and optimum choices*, Pol. Przegl. Chir., 2018, 90 (2), 34–37, DOI: 10.5604/01.3001.0010.5632.

[14] GIEREK M., MERKEL K., OCHAŁA-GIEREK G., NIEMIEC P., SZYLUK K., KUŚNIERZ K., *Which Suture to Choose in Hepato-Pancreatic-Biliary Surgery? Assessment of the Influence of Pancreatic Juice and Bile on the Resistance of Suturing Materials – In Vitro Research*, Biomedicines, 2022, 10 (5), 1053, DOI: 10.3390/biomedicines10051053.

[15] GUAMBO M.P.R., SPENCER L., VISPO N.S., VIZUETE K., DEBUT A., WHITEHEAD D.C., SANTOS-OLIVEIRA R., ALEXIS F., *Natural Cellulose Fibers for Surgical Suture Applications*, Polymers, 2020, 12 (12), 3042, DOI: 10.3390/polym12123042.

[16] GUPTA V., *Potential of Natural Plant-Based Materials in the Development of Biocompatible Drug-Eluting Surgical Sutures: A Review*, Biomed. Mater. Devices, 2025, 3, 1125–1149, DOI: 10.1007/s44174-024-00259-0.

[17] HAKIM M.L., NAHAR N., SAHA M., ISLAM M.S., REZA H.M., SHARKER S.M., *Local drug delivery from surgical thread for area-specific anesthesia*, Biomed. Phys. Eng. Express, 2020, 6 (1), 015028, DOI: 10.1088/2057-1976/ab6a1e.

[18] HASSAN H.K., *Dental Suturing Materials and Techniques*, Glob. J. Otolaryngol., 2017, 12 (2), 555833, DOI: 10.19080/GJO.2017.12.555833.

[19] Jinhuan Medical Absorbable surgical suture product (trade name (PGA)) http://en.sh-jinhuan.com/f2c61b8d-0977-6255-721d-62dfc7705318/08152ba2-f938-3229-fd16-8d1004fa8341.shtml [Accessed: 30.07.2025]

[20] KAMPEERAPAPPUN P., DESCLAUX S., RATTANAPINYOPITUK K. et al., *Natural cellulose fibers derived from Dracaena angolensis (Welw. ex Carrière) Byng & Christenh. demonstrate potential as a non-absorbable surgical suture biomaterial*, Sci. Rep., 2025, 15, 1291, DOI: 10.1038/s41598-025-85886-3.

[21] KARPIŃSKI R., SZABELSKI J., MAKSYMIUK J., *Effect of Ringer's Solution on Tensile Strength of Non-Absorbable, Medium- and Long-Term Absorbable Sutures*, Adv. Sci. Technol. Res. J., 2017, 11 (4), 11–20, DOI: 10.12913/22998624/76084.

[22] KASHIWABUCHI F., PARIKH K.S., OMIADZE R., ZHANG S., LUO L., PATEL H.V., XU Q., ENSIGN L.M., MAO H.Q., HANES J., MCDONNELL P.J., *Development of Absorbable, Antibiotic--Eluting Sutures for Ophthalmic Surgery*, Transl. Vis. Sci. Technol., 2017, 6 (1), 1, DOI: 10.1167/tvst.6.1.1.

[23] KAWAI F., KAWABATA T., ODA M., *Current knowledge on enzymatic PET degradation and its possible application to waste stream management and other fields*, Appl. Microbiol. Biotechnol., 2019, 103, 4253–4268, DOI: 10.1007/s00253-019-09717-y.

[24] KUDUR M.H., PAI S.B., SRIPATHI H., PRABHU S., *Sutures and suturing techniques in skin closure*, Indian J. Dermatol. Venereol. Leprol., 2009, 75 (4), 425–434, DOI: 10.4103/0378-6323.53155.

[25] LI Y., MENG Q., CHEN S., LING P., KUSS M.A., DUAN B., WU S., *Advances, challenges, and prospects for surgical suture materials*, Acta Biomater., 2023, 168, 78–112, DOI: 10.1016/j.actbio.2023.07.041.

[26] MADSEN K., MARTENS A., HASPESLAGH M., MEULYZER M., GUSTAFSSON K., *The effect of medical grade honey on tensile strength, strain, and Young's modulus of synthetic absorbable suture material used in equine surgery*, Equine Vet. J., 2024, 56 (1), 193–201, DOI: 10.1111/evj.13966.

[27] MATALIA J., PANMAND P., GHALLA P., *Comparative analysis of non-absorbable 10-0 nylon sutures with absorbable 10-0 Vicryl sutures in pediatric cataract surgery*, Indian J. Ophthalmol., 2018, 66 (5), 661–664, DOI: 10.4103/ijo.IJO_654_17.

[28] MERKEL K., GRZYBOWSKA K., STRACH A., GIEREK M., *The Degradation of Absorbable Surgical Threads in Body Fluids: Insights from Infrared Spectroscopy Studies*, Int. J. Mol. Sci. 2024, 25 (20), 11333, DOI: 10.3390/ijms252011333.

[29] MÜLLER D.A., SNEDEKER J.G., MEYER D.C., *Two-month longitudinal study of mechanical properties of absorbable sutures used in orthopedic surgery*, J. Orthop. Surg. Res., 2016, 11, 111, DOI: 10.1186/s13018-016-0451-5.

[30] NICULESCU M., ANTONIAC A., VASILE E., SEMENESCU A., TRANTE O., SOHACIU M., MUSETESCU A., *Evaluation of biodegradability of surgical synthetic absorbable suture materials: An in vitro study*, Mater. Plast., 2016, 53 (4), 642–645.

[31] OJASTHA B. L., JEEVITHA M., *An Evaluation of the Tensile Strength of Polyglactin Sutures After Immersion in Different Herbal Mouthwashes: An In Vitro Study*, Cureus, 2023, 15 (8), e43407, DOI: 10.7759/cureus.43407.

[32] OOI, C.P., CAMERON R.E., *The hydrolytic degradation of polydioxanone (PDSII) sutures*. Part I: *Morphological aspects*, J. Biomed. Mater. Res., 2002, 63 (3), 280–290, DOI: 10.1002/jbm.10180.

[33] ROSE J., TUMA F., *Sutures And Needles*, 2023 Aug. 28. In: StatPearls [Internet]. Treasure Island (FL): StatPearls Publishing; 2025 Jan. Available from: https://www.ncbi.nlm.nih.gov/books/NBK539891/

[34] SHAWE S., BUCHANAN F., HARKIN-JONES E. et al., *A study on the rate of degradation of the bioabsorbable polymer polyglycolic acid (PGA)*, J Mater. Sci., 2006, 41, 4832–4838, DOI: 10.1007/s10853-006-0064-1.

[35] SHENG S.T., WU X.D., XU J.W., XU Z., NI S., XU W., XU Z.K., *Biomaterials in Relative Devices for Traumatic Cataract: Recent Advances and Future Perspectives*, ACS Biomater. Sci. Eng., 2025, 11 (4), 1990−2007, DOI: 10.1021/acsbiomaterials.4c02117.

[36] SRIYAI M., TASATI J., MOLLOY R., MEEPOWPAN P., SOMSUNAN R., WORAJITTIPHON P., DARANARONG D., MEERAK J., PUNYODOM W., *Development of an Antimicrobial-Coated Absorbable Monofilament Suture from a Medical-Grade Poly(l-lactide-co-ε-caprolactone) Copolymer*, ACS Omega. 2021, 6 (43), 28788–28803, DOI: 10.1021/acsomega.1c03569.

[37] SZABELSKI J., KARPIŃSKI R., *Short-Term Hydrolytic Degradation of Mechanical Properties of Absorbable Surgical Sutures: A Comparative Study*, J. Funct. Biomater., 2024, 15 (9), 273, DOI: 10.3390/jfb15090273.

[38] TUMMALAPALLI M. et al., *Antimicrobial Surgical Sutures: Recent Developments and Strategies*, Polym. Rev., 2016, 56 (4), 607–630, DOI: 10.1080/15583724.2015.1119163.

[39] TURAN S.K., KILIÇ SÜLOĞLU A., İDE S., TÜRKEŞ T., BARLAS N., *In vitro and in vivo investigation of Argiope bruennichi spider silk-based novel biomaterial for medical use*, Biopolymers, 2024, 115 (3), e23572, DOI: 10.1002/bip.23572.

[40] VOGELS R.R., LAMBERTZ A., SCHUSTER P., JOCKENHOEVEL S., BOUVY N.D., DISSELHORST-KLUG C., NEUMANN U.P., KLINGE U., KLINK C.D., *Biocompatibility and biomechanical analysis of elastic TPU threads as new suture material*, J. Biomed. Mater. Res. B., Appl. Biomater., 2017, 105 (1), 99–106, DOI: 10.1002/jbm.b.33531.

[41] YANG Z., LIU S., LI J. et al., *Study on Preparation of Core-Spun Yarn Surgical Sutures by Compositing Drug-Loaded Nanofiber Membrane with PLA and Its Controllable Drug Release Performance*, Fibers Polym., 2023, 24, 4181–4193, DOI: 10.1007/s12221-023-00386-3.

[42] ZUREK M., KAJZER A., BASIAGA M., JENDRUŚ R., *Właściwości wytrzymałościowe wybranych polimerowych nici chirurgicznych*, Polimery, 2016, 61 (5), 334–338, DOI: 10.14314/polimery.2016.334.