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The Stress Relaxation Process in Sutures Tied with a Surgeon's Knot in a Simulated Biological Environment

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A – research concept and design; **B** – collection and/or assembly of data; **C** – data analysis and interpretation;
D – writing the article; **E** – critical revision of the article; **F** – final approval of the article

Abstract

Background. The exact characteristics of sutures are not only the basis for selecting from among different types of suture, but also provide the necessary information for the design of new surgical sutures. Apart from information relating to the breaking load of a suture reported in pharmacopoeias, the viscoelastic properties of sutures can be an additional selection criterium – one that influences stitching quality, especially when there is a risk of wound dehiscence.
Objectives. The aim of the study was to assess the stress relaxation process for 3 polymeric sutures in an environment simulating the conditions in a living organism and (for comparison) in room conditions.
Material and Methods. Stress relaxation testing was carried out on 3 polymeric sutures: polypropylene (PP), polydioxanone (PDS) and polyglycolic acid (PGA). To identify the mechanical properties of the sutures, uniaxial tensile tests were conducted according to the Polish Pharmacopoeia. The relaxation test was carried out in room conditions and in the bath simulating a biological environment. The sutures being tested were tied with a surgeon's knot.
Results. The PP suture exhibited the greatest stress relaxation (18% of the initial stress in room conditions and 21% of the initial stress in the bath). The PGA suture exhibited the least stress relaxation (approximately 60% of the initial stress in room conditions and 59% of the initial stress in the bath). The PDS suture was tested at a lower strain level and showed stress relaxation similar to the PGA suture (approximately 63% of the initial stress in room conditions and 55% in the bath).
Conclusions. Multifilament braided absorbable (PGA) sutures and monofilament absorbable (PDS) sutures had a higher stress relaxation ratio over time than monofilament non-absorbable (PP) sutures. These findings may indicate higher stress maintained over time in PDS and PGA sutures, and thus higher tension at wound edges, sufficient to resist wound dehiscence (Polim. Med. 2016, **46**, 2, **111–116**).

Key words: mechanical properties, stress relaxation, residual stress, polymeric sutures, surgeon's knot.

There are many types of sutures on the medical devices market, classified according to their origin (natural/synthetic), structure (monofilament/multifilament, braided or twisted) and the time they remain in tissues (absorbable/non-absorbable). As medical devices, surgical sutures must meet the essential requirements of the Polish Pharmacopoeia, which incorporates the standards of the European Pharmacopoeia, including physical properties such as diameter, break load of the suture tied into a single knot, the durability of the needle-suture connection, packaging and sterility [1]. Suture selection should be based on knowledge of the physical and biological characteristics of the material in relation to the healing process. The surgeon wants to ensure that a suture will retain its strength until the tissue has healed sufficiently to withstand normal stress. Evaluations of surgical sutures have been the subject of many studies, mainly related to the break load of sutures, the properties of different surgical knots, the degree of absorption of the suture, tissue response and wound infection [2–6].

The exact characteristics of sutures are not only the basis for selecting the type of suture, but also provide necessary information for the design of new surgical sutures. Apart from information relating to the break load of the suture reported in the Pharmacopoeia, the viscoelastic properties of sutures can be an additional selection criterium – one that influences stitching quality, especially when there is a risk of wound dehiscence. Relatively few studies on the viscoelastic properties of sutures can be found in the available literature. It must also be borne in mind that due to the large variety of sutures types differing in size, material, structure, knot types and retention in tissues, test results are difficult to compare.

This paper presents the results of stress relaxation tests for 3 polymeric sutures in an environment simulating the conditions of a living organism and, for comparison, in room conditions. The sutures tested differed in material, structure and retention in tissues.

Material and Methods

The material used in the study comprised 3 size 2 sutures (diameter 0.2 mm) according to the sizing used in the Polish Pharmacopoeia:

– polypropylene (PP): Surgipro non-absorbable monofilament (Medtronic, Minneapolis, USA),

– polydioxanone (PDS): Surgicryl absorbable monofilament (SMI AG, Vith, Belgium),

– polyglycolic acid (PGA): Polycryl absorbable braided multifilament (Aurolab, Veerapanjan, India).

To determine the mechanical properties of the sutures, uniaxial tensile tests were conducted according to the Polish Pharmacopoeia using an Insight 50 tensile machine (MTS Systems Corporation, Eden Prairie, USA) with a load cell capacity of 1 kN. The gauge length (the test section of the sutures) was 125 mm. A simple knot was tied in the middle of the test section of each suture. At least 5 specimens of each type were tested, and the speed of elongation was 50 mm/min. The average breaking force was obtained and compared with the requirements of the Polish Pharmacopoeia, and the breaking stress was calculated as:

$$\sigma_B = \frac{F_B}{A}[MPa], \tag{1}$$

where F_B (N) is the breaking force and A (mm^2) is the cross sectional area of the suture.

Stress relaxation is generally defined as the stress change in a material subjected to a constant strain at a constant temperature. The stress relaxation experiments were conducted on the Insight 50 tensile machine described above. The suture was connected to the tensile machine with suture grips (Fig. 1). A surgical knot was tied in the middle of the test section of each suture (Fig. 2).

The relaxation tests were carried out in room conditions (23 ± 1°C, 65% RH) and in simulated biological environment conditions (0.9% NaCl solution at a temperature of 37 ± 1°C). They were performed with the use of an attaching system in a temperature-controlled saline solution bath. The suture was connected to the Insight 50 tensile machine by suture grips and completely immersed in the bath (Fig. 3). While in the grips, the suture samples were placed in the saline solution bath at the test temperature for about 10 min to reach equilibrium before a load was applied. The strain

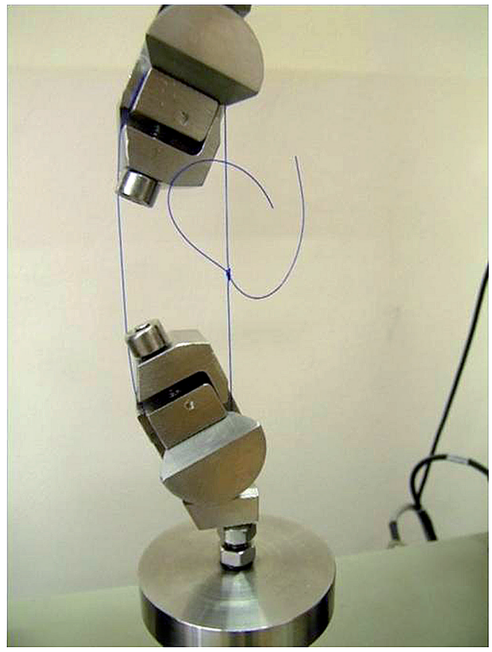

Fig. 1. A suture sample in the grip during the tensile test

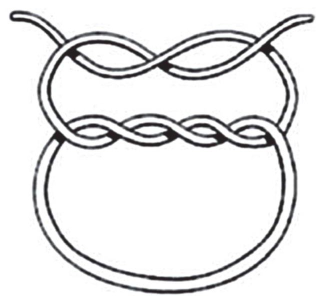

Fig. 2. The type of surgical knot used in the relaxation tests [6]

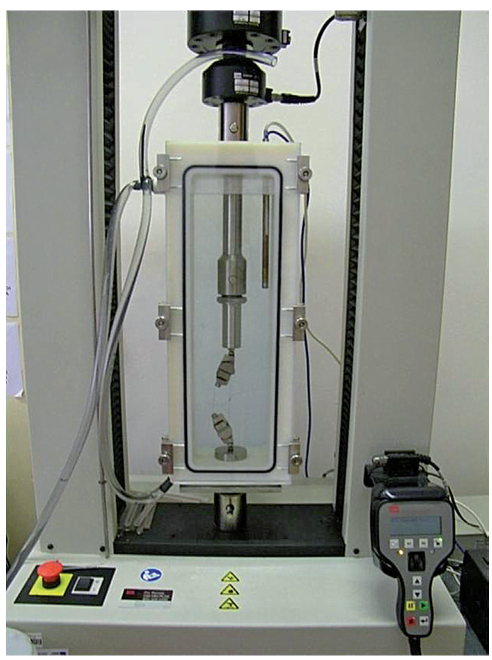

Fig. 3. A suture sample in the grip in the saline solution bath during the relaxation test

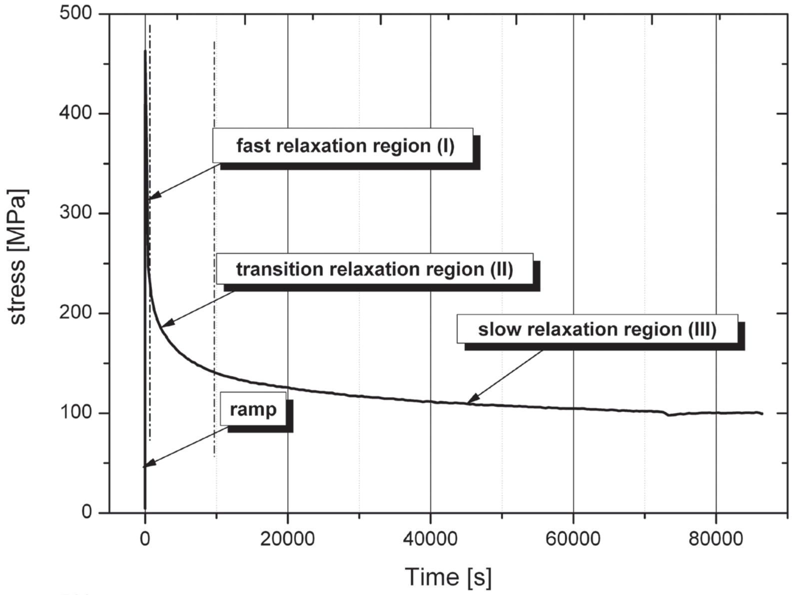

Fig. 4. Three characteristic regions of true stress relaxation curves: the fast relaxation region (I), the transition region (II) and the slow relaxation region (III)

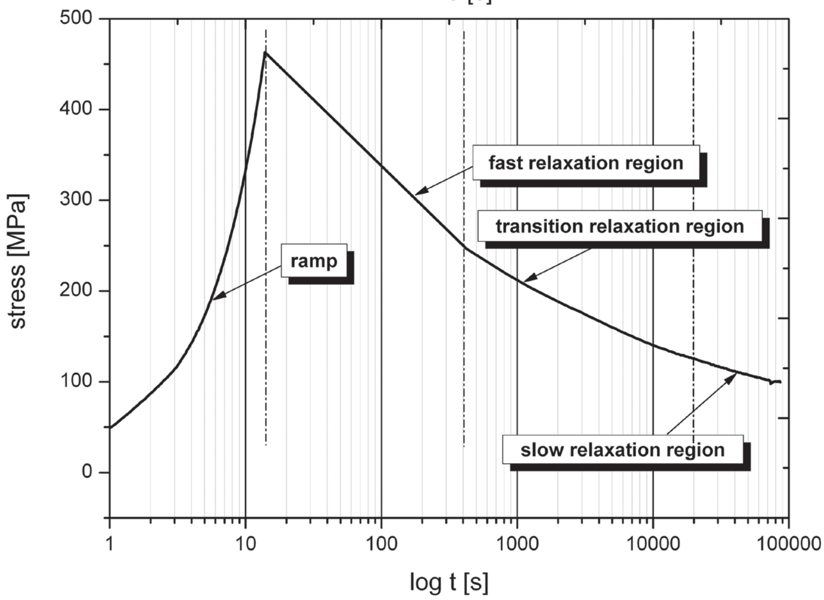

Fig. 5. Three characteristic regions of true stress relaxation curves (on a logarithmic scale of time): the fast relaxation region (I), the transition region (II) and the slow relaxation region (III)

level for the stress relaxation tests was 50%. For the PDS suture the selected strain level turned out to be too high (during the ramping phase the suture broke at the knot), so the strain level was reduced to 23%. Each stress relaxation experiment lasted 24 h.

For each set of conditions, at least 2 specimens were tested. The load, time and elongation were recorded, and changes in stress over time during the stress relaxation tests were calculated. Three characteristic ranges can be distinguished on the stress relaxation curve (Figs. 4, 5). In the initial phase of relaxation, the fast relaxation range can be observed; then the second range (the transition range) and the slow relaxation range (the longest) can be seen. The stress relaxation ratio (G(t), normalized stress) was calculated as:

$$g(t) = stress\ relaxation\ ratio = \frac{\sigma_i}{\sigma_0} \qquad (2)$$

where, σ_i is stress at time, t = (0,∞) (here 24 h) and σ_0 is the initial stress at a constant level of strain.

Results

The load at break obtained in the tensile test for the sutures tested is presented in Fig. 6. The polydioxanone suture had the highest load at break values (24.1 ± 2.1 N) and breaking stress (767.52 ± 66.88 MPa). The polyglycolide suture had lower breaking load values (20.2 ± 0.3 N) and breaking stress (643.31 ± 9.55 MPa), followed by the polypropylene suture (breaking load value: 13.2 ± 1.5 N, breaking stress value: 420.38 ± 47.77 MPa). The breaking load values for all the sutures tested were higher than the minimum required values according to the Polish Pharmacopoeia [1].

The results of the stress relaxation tests are presented in Figs. 7–9. It can be seen that in all the sutures stress decreased over time. Three regions can be seen in the stress relaxation curves: the initial phase of relaxation is the fast relaxation region; the slow relaxation region is the longest; and the region between these two

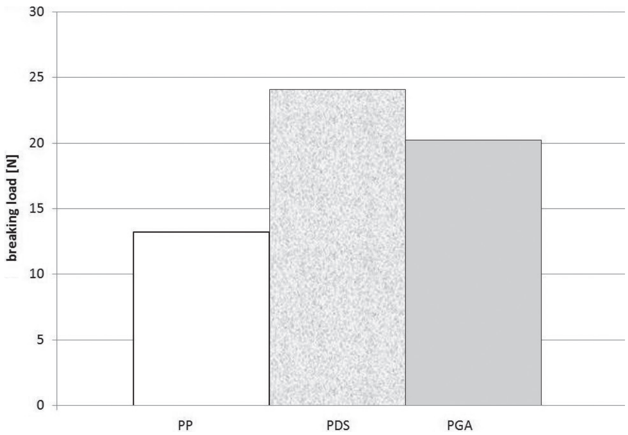

Fig. 6. The average breaking load for the sutures tested

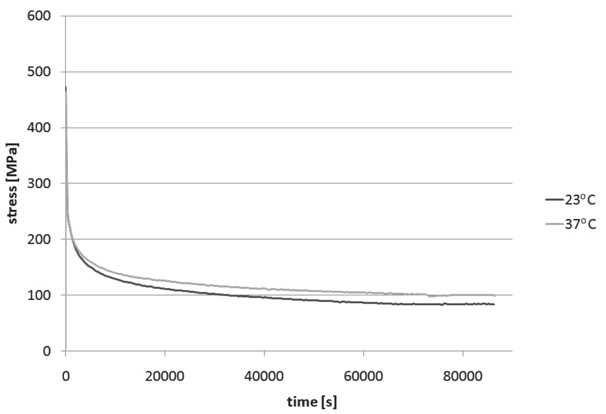

Fig. 7. A comparison of the relaxation curves for the polypropylene suture in room conditions (23°C) and in the simulated biological environment (37°C)

is the transition region. The fast relaxation phase ended very quickly for all the sutures tested, lasting about 400 s. The transition phase was completed in less than 7 min. In room conditions the transition phase lasted about 2000 s for all the specimens; in simulated biological conditions it lasted about 2000 s for the PP and PDS sutures, and about 800 s for the PGA suture. In room conditions the slow relaxation phase started at about 10,000 s for the PP suture, 6500 s for the PDS suture and 5500 s for the PGA suture. This phase of relaxation started earlier in the simulated biological environment: at 1900 s for the PP suture; at 5000 s for the PDS suture; and at 3000 s for the PGA suture. The stress in the sutures decreased gradually in this phase.

A significant difference in stress relaxation and residual stress values was observed between polypropylene and polyglycolic acid sutures. The residual stress ratio for polypropylene was approximately 40% lower than polyglycolic acid. This difference may result from the structure of the sutures: the PGA suture was a multifilament suture and therefore had better knot security.

The polypropylene suture exhibited the greatest stress relaxation. The residual stress value was approximately 18% of the initial stress in room conditions and 21% of the initial stress in the bath after 24 h of the test (Table 1). The PGA suture (at the same strain level) exhibited the lowest stress relaxation of the 3 suture types tested: the residual stress was approximately 60% of the initial stress in room conditions and 59% of the initial stress in the bath. The PDS suture tested at a lower strain level showed similar stress relaxation to the PGA suture: the residual stress was approximately 63% of the initial stress in room conditions and 55% in the bath.

Table 1. Residual stress values after 24 h as percentages of the initial stress

Suture	Room conditions	Simulated biological conditions
PP	18%	21%
PDS	63%	55%
PGA	60%	59%

The test conditions influenced the stress relaxation process of the PDS suture (Fig. 8). For this suture, a significantly lower residual stress value was observed in conditions simulating a biological environment in comparison to the residual stress value in room conditions. For the PP (Fig. 7) and PGA (Fig. 9) sutures, there was only a slight difference between the relaxation curves obtained in the 2 test conditions, and it was within the margin of error.

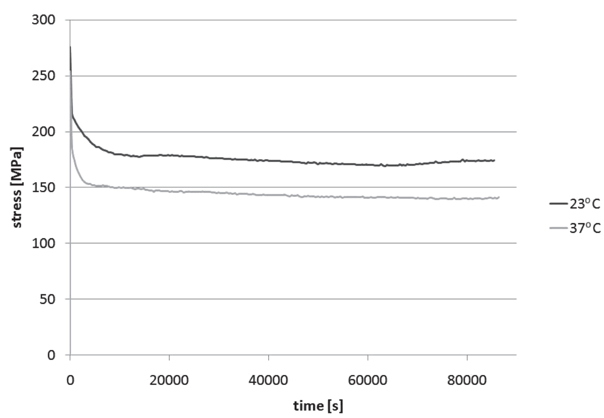

Fig. 8. A comparison of the relaxation curves for the polydioxanone suture in room conditions (23°C) and in the simulated biological environment (37°C)

In Fig. 10, the stress relaxation ratio over time in room conditions is presented. After 24 h of the relaxation process, the value of the stress relaxation ratio was 0.18 ± 0.01 for the PP suture; 0.63 ± 0.01 for the PDS suture; and 0.60 ± 0.01 for the PGA suture. The conditions simulating a biological environment caused a decrease in the stress relaxation ratio for the PDS suture (0.55 ± 0.01). A slight decrease was observed for the PGA suture (0.59 ± 0.01) and a slight increase for the

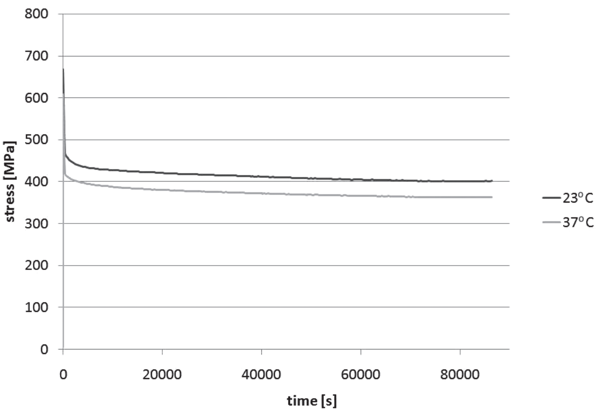

Fig. 9. A comparison of the relaxation curves for the polyglycolide suture in room conditions (23°C) and in the simulated biological environment (37°C)

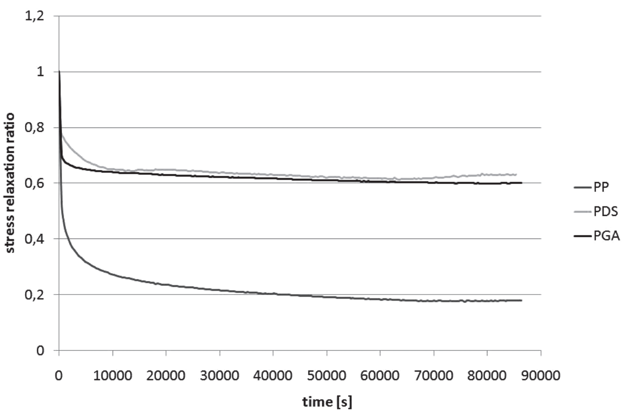

Fig. 10. The stress relaxation ratio for the sutures tested in room conditions

PP suture (0.21 ± 0.01), but these differences were within the margin of error (Fig. 11). The stress value after 24 h of the relaxation process in relation to the ultimate tensile strength (breaking stress) was about 20% lower for the PP suture, 23% for the PDS suture and 62% for the PGA suture.

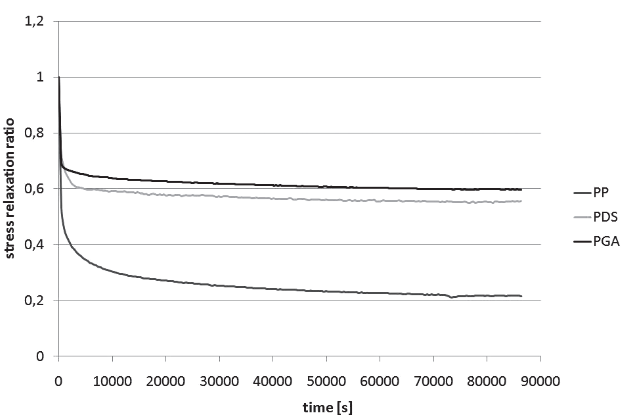

Fig. 11. The stress relaxation ratio for the sutures tested in simulated biological conditions

Discussion

In the present study the greatest stress relaxation was observed in the polypropylene suture, which concurs with the results presented by Vizesi et al. and Metz et al. [7, 8]. According to Vizesi et al. the polypropylene suture showed significantly higher stress relaxation than nylon and silicone-coated polyester, and was the only suture dependent on the temperature of the test [7]. All the sutures tested by Metz et al. (PP, PGA and PDS) exhibited stress relaxation to different degrees, with polypropylene exhibiting the greatest stress relaxation. The initial rate of stress relaxation increased with the loading rate, but the residual stress level was found to be independent of the elongation rate except at extremely low loading rates [8]. In the present study, the PP stress ratio in the 10th min of the test was calculated at approximately 0.47, which was in good agreement with Vizesi et al. (approximately 0.42) [7].

Deng et al. evaluated the stress relaxation of polypropylene sutures and concluded that the strain level and the temperature had significant effects on the stress relaxation properties of the sutures. Increasing the strain level in the PP suture relaxation test caused higher stress, and increasing the temperature resulted in lower stress [9]. The significant influence of temperature on PP relaxation was observed by Deng et al. at all strain levels (13, 26 and 33%) over 10,000 s, and by Vizesi et al. at 10% strain over 600 s. The difference between the present results and those of these authors may result from the differences in suture size, strain levels and the longer duration of the relaxation process, as well as the air conditions at room temperature [7, 9].

Considering the problem of stress relaxation in sutures, it is necessary to compare the results with the properties of soft tissues. A search of the available literature showed that skin tissue analyzed for stress relaxation properties has similar behavior to polypropylene sutures. According to Liu and Yeung, normalized stress (the stress relaxation ratio) for pig skin tissue reached a value between 0.2 and 0.6 after 1200 s of the stress relaxation test depending on the direction of the sample taken (soft tissue anisotropy) and the level of strain [10]. A stress relaxation ratio of 0.2 for pig skin tissue after 1800–4000 s of the stress relaxation process, depending on the level of strain and the direction of the sample taken was reported by Liber-Kneć [11].

Conclusions

The stress relaxation behavior of 3 polymeric sutures was investigated. Additionally, the influence of the relaxation test conditions was assessed. Manually tying and tensioning a surgical knot is easier in the case of a multifilament suture (convenience suture), although it may limit the stress relaxation. Multifilament braided absorbable PGA and monofilament absorbable PDS sutures had higher stress relaxation ratio values over time than the monofilament non-absorbable PP suture. These findings may indicate a higher level of stress remaining over time in PDS and PGA sutures, and thus higher tension at wound edges, sufficient to resist wound dehiscence. To characterize the detailed viscoelastic properties of suture materials, further investigation is needed, taking into consideration the structure of the sutures and the types of surgical knots.

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