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# Stress relaxation tests in polypropylene monofilament meshes used in the repair of abdominal walls

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The objective of this work has been to characterize stress relaxation in the polymer material on applying different levels of constant strain. The meshes were strained at values of 5.2%, 5.4% and 5.6% which are the values at which the mesh is strained in clinical use for the repair of abdominal walls. Laws have been obtained to model the viscoelastic behavior at different strains for this material. Finally, fracture studies were carried out by environmental scanning electron microscopy to determine the fracture mechanisms of these meshes. Besides, the implantation of the meshes was practised in two different layers of abdominal wall: the superficial or preaponeurotic layer and deep or preperitoneal layer, showing the neoformation of connective tissue on the mesh, which tended to be organized differently in each layer studied; more roughly and densely in the superficial layer than in the deep one.


## Introduction

Eventration has always been a problem that is difficult to solve from the surgicomedical point of view. The fact that different procedures are described for surgical treatment shows that there is none, which is so superior to the others as to supersede them, or that all of them, from the personal or medical school point of view, have more or less satisfactorily solved the problem [1]. The procedure must not be thought of as ''closing'' an orifice in the wall, especially in large eventrations, but as re-establishing the functions of the abdominal wall, and adequately dealing with the viscera contained in the sac. The abdominal muscles must recover their previous pressure points and they must be put under an average stress to reestablish intra-abdominal pressure that in normal conditions is 6–8 mm Hg above the diaphragm and 12 mm Hg at the level of the pelvis, achieving a solid wall and the best possible appearance [2–5].

The mechanical behavior of polypropylene is viscoelastic and it therefore presents phenomena of flow and stress relaxation that will be influenced by the temperature of the mechanical test; undoubtedly, a temperature of 37 °C accelerates the processes of stress relaxation. Polypropylene is a highly rigid polyolefine with high chemical and thermal resistance and good mechanical properties; in particular, it has excellent bending behavior over long periods of time [5–9].

In this work a study has been carried out of the stress relaxation of polypropylene (Marlex®) monofilament meshes using high strain values at 37 °C in physiological serum to simulate as far as possible the environment in which these meshes are used.

## Material and experimental method

Meshes were implanted in a total of twelve 16-week-old male Sprague–Dawley rats weighing between 350 and 400 g. The strips contained the interfaces formed with the on-lay technique, including the prosthesis (P) and the subcutaneous tissue layer and recipient muscle aponeurosis (superficial stratum: SM); and the interfaces formed with the in-lay technique, including the prosthesis and the transverse abdominal muscle and the parietal peritoneal surface (deep stratum: P-DM). The size of the meshes was $3 \times 3\ cm^2$. Studies were performed over implantation intervals of 7, 15, and 30 days. At sacrifice, the anterior abdominal wall was resected and samples were taken parallel to the shorter axis of the implant. At each time the animals were killed and the samples were extracted to be observed by electron microscopy for the carrying out of the mechanical tests.

An Electroscan 2020 environmental scanning electron microscope (ESEM) equipped with a Peltier effect sample holder was used to allow the observation of the samples in their natural state without the need for previous preparation.

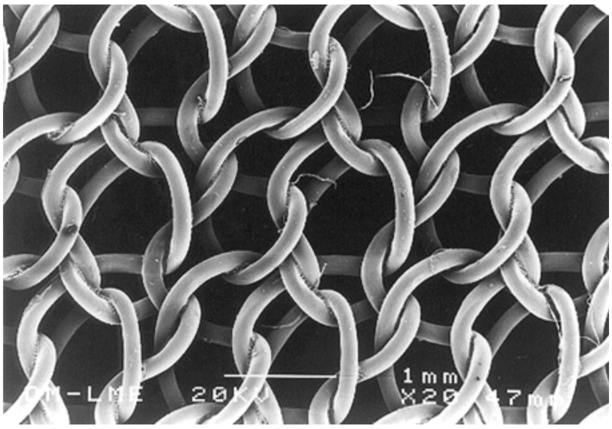


*Figure 1* Polypropylene monofilament mesh.

Five samples of polypropylene monofilament mesh were studied, as can be seen in Fig. 1. In Fig. 2, the meshes show points of adhesion, which, in general, improve their mechanical properties. This configuration gives the mesh good elasticity with a high level of mechanical strength. The entirety of the mesh contracts when one of the fibers breaks, as the whole structure contributes towards sharing the mechanical load.

These meshes were strained at values of 5.2%, 5.4% and 5.6%, which are the values at which the mesh is strained in clinical use for the repair of abdominal walls [10]. The mechanical tests were carried out with an MTS-Adamel electromechanical testing machine. The rate of testing was 10 mm/min and a load cell of 1 kN was used because it has greater sensitivity in the force values that must be applied to maintain these levels of strain. The force values were determined at different times. To control the constant strain a laser extensometer was used. This type of extensometer was used with the aim that the pressure should not cause cuts in the polymer and therefore affect the values of the mechanical forces measured. With the laser extensometer fluorescent tapes were placed on the sample to be measured in order to allow the laser to measure the displacement values remotely.

With the aim of carrying out the mechanical tests in conditions similar to the human body, a physiological chamber was designed at a temperature of 37 °C. The rate of application of the load to arrive at the strain studied was 1 mm/min and the rate of data acquisition by the Autotrack software was 25 points/second; this enabled us to control automatically the stress relaxation that is produced over time.

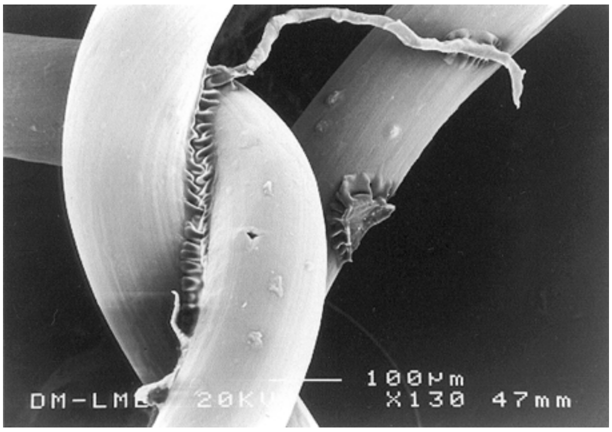


*Figure 2* Points of adhesion of the meshes.

## Experimental results and discussion

From the studies carried out, ESEM showed the neoformation of connective tissue on the mesh, which tended to be organized differently in each layer studied; more roughly and densely in the superficial layer than in the deep one, as can be seen in Figs. 3 and 4, respectively. This difference was more clearly visible in the anchorage zones of the host tissue. After day 15 following the operation a more mature and sclerous extracellular component could be seen in the superficial layer. In the deep layer a more oedematose tissue predominated and after 30 days from the implantation of the meshes the

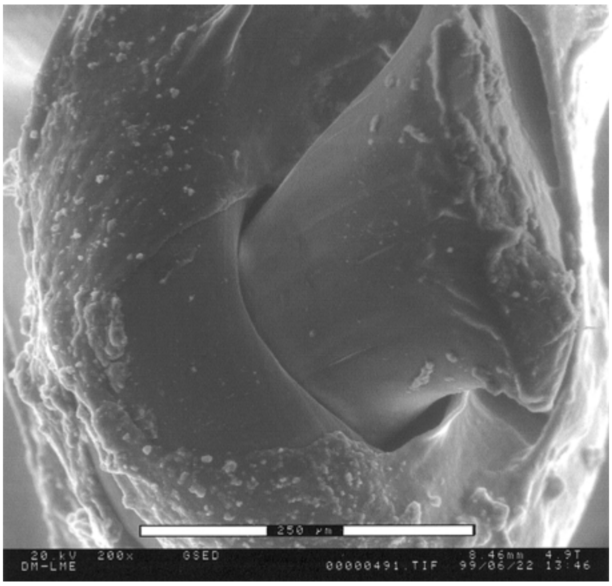

*Figure 3* Neoformation of connective tissue on the mesh on the superficial layer.

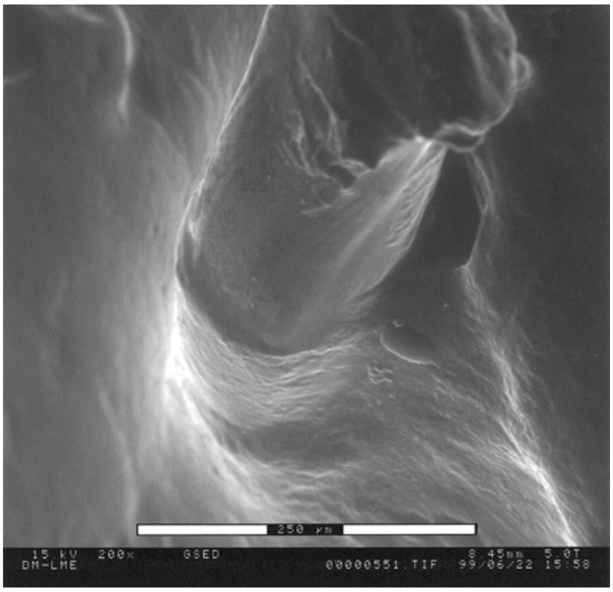

*Figure 4* Neoformation of connective tissue on the mesh on the deep layer.

TABLE I Effect of placement and interval post-implantation on tensiometric measurements

| | P–SS interface | | | P–DS interface | | | RSE | Variance analysis, significant effects | | |
|---|---|---|---|---|---|---|---|---|---|---|
| | 7 d | 15 d | 30 d | 7 d | 15 d | 30 d | | P | I | PI |
| *n* | 3 | 5 | 4 | 3 | 5 | 4 | | | | |
| Tension strength, N/mm$^2$ | 0.220 | 0.214 | 0.320 | 0.163 | 0.158 | 0.212 | 0.014 | 0.016 | ns | ns |
| Elongation (%) | 24.660 | 23.000 | 35.970 | 20.330 | 21.400 | 34.025 | 1.510 | ns | 0.003 | ns |
| Stiffness, N/mm | 0.510 | 0.508 | 0.505 | 0.490 | 0.502 | 0.510 | 0.004 | ns | ns | ns |

Values are expressed as mean and residual SE (RSE); *n* refers to no. rats/group. 12-wk Sprague–Dawley rats received two pre-cut graft implants using an *on-lay* technique (mesh under the subcutaneous fat and on the abdominal aponeurotic layer) and an *in-lay* technique (mesh on the peritoneal surface). Two-way variance analyses were performed to discriminate among effects of placement (P) and interval post-implantation (I), and their interaction (PI) on tensiometric studies; ns = non-significant.

cellurarity was lower in the deep layer than in the superficial layer.

The SM series offered a better tensile strength than the DM series. In this way it can be shown that as we lengthen the time of insertion, the mechanical stress increases due to the improvement in adhesion between the mesh and the tissue. The differences can be seen in Table I for the different times studied. Two-way variance analysis demonstrated that the post-implantation interval had no effect on tensile strength measurements during the period studied as can be observed in Table II and the effect of implantation intervals on percent elongation in Table III.

It has been possible to determine the strain values for the polypropylene meshes at the moment of maximum resistance of the tissue–mesh interface for each of the studies carried out, varying from 4.8% to 6.4%, without significant differences being noticed, in the ANOVA test with a $p < 0.005$, compared to total strain values for the sample which oscillated between values of 21% and 35%. On the basis of these results, experiments were designed to relax the forces for the maximum strain values the meshes could reach. These were as stated in the section Material and experimental method: 5.2%, 5.4% and 5.6% of total strain.

Figs 5–7 show the curves for stress relaxation at strains of 5.2%, 5.4% and 5.6%. In view of the results at the different levels of strain, it can be seen that the same asymptotic rule of stress relaxation is produced. As was to be expected, to achieve a level of strain of 5.6%, a force of 3.58 N is needed, which is the maximum for the different strains, and to achieve the smallest strain (5.2%), a stress of 2.3 N is needed, which is the minimum.

The stress saturation levels – i.e. the points at which the stress becomes practically constant with time – are 1.5 N for strain levels of 5.2 and 5.4% and above 2 N for the 5.6% strain level. Note the great speed with which

TABLE II Effect of placement position on tensile strength

| Placement | *n* | N/mm$^2$ |
|---|---|---|
| P–SS interface | 12 | 0.251 ± 0.023* |
| P–DS interface | 12 | 0.177 ± 0.018 |

Two-way variance analysis demonstrated that the post-implantation interval had no effect on tensile strength measurements during the period studied (see Table I). Thus, values obtained at 7, 15 and 30 days were pooled for each interface considered. Values are expressed as mean ± SEM; *n* refers to no. rats/group.
*$P < 0.02$.

TABLE III Effect of implantation intervals on percent elongation

| Days post-implantation | *n* | Percent elongation |
|---|---|---|
| 7 | 6 | 19.2 ± 1.4 |
| 15 | 10 | 18.6 ± 2.0 |
| 30 | 8 | 26.7 ± 3.6* |

Two-way variance analysis demonstrated that the placement technique had no effect on percent elongation measurements during the period studied (see Table I). Thus, values obtained in polypropylene fiber-superficial and -deep stratum interfaces were pooled at each post-implantation interval (7, 15 and 30 days). Values are expressed as mean ± SEM; *n* refers to no. rats/group.
*$P \leq 0.005$ vs. 7 and 15 days.

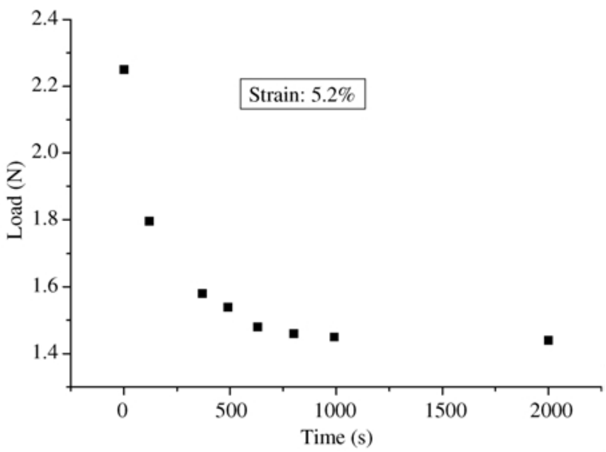


*Figure 5* Stress relaxation at strains of 5.2%.

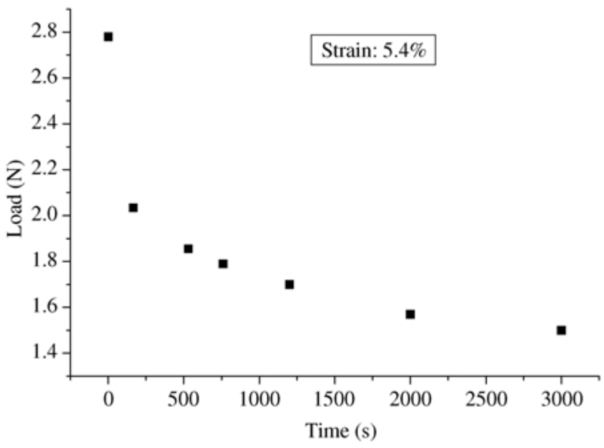


*Figure 6* Stress relaxation at strains of 5.4%.

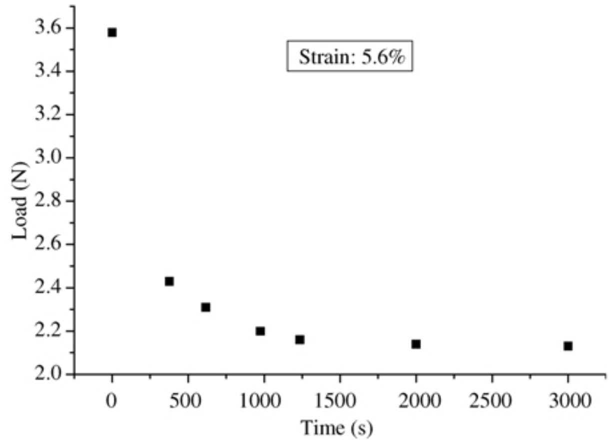


*Figure 7* Stress relaxation at strains of 5.6%.

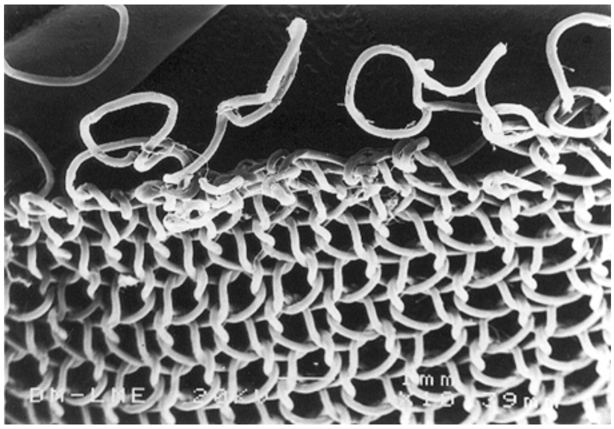

*Figure 8* Contraction of the whole mesh in the area of the fracture.

these materials relax their stresses and reach constant force values. This is a positive factor in that clinics will know that the force to achieve a certain strain decreases in a short time of approximately 500 s. From this time the level of mechanical stress to which the polypropylene mesh is being submitted can be assured.

These relaxation curves can be adjusted according to the following equation:

$$\mathrm{Ln}\ \sigma = \mathrm{Ln}\ \sigma_0 - B(t/\tau)$$

where $\sigma$: stress as a function of time; $\sigma_0$: stress at time zero; $t$: time and $\tau$: relaxation time.

By adjusting the equation with experimental data, laws to model the viscoelastic behavior of these materials at different strains can be obtained.

Strain 5.2% $\quad \sigma_0 = 0.146\,\mathrm{PMa}$, $\tau = 3401.4\,\mathrm{s}^{-1}$, $B = -1.925$

Strain 5.4% $\quad \sigma_0 = 0.162\,\mathrm{PMa}$, $\tau = 3496.5\,\mathrm{s}^{-1}$, $B = -1.822$

Strain 5.6% $\quad \sigma_0 = 0.205\,\mathrm{PMa}$, $\tau = 3381.8\,\mathrm{s}^{-1}$, $B = -1.585$

These equations will allow the clinic to be able to predict how the mesh forces relax with the passing of time for a certain strain applied.

Fractographic studies show that a rupture in the monofilament mesh begins in one fiber and there is a contraction of the whole mesh in the area of the fracture as can be observed in Fig. 8. It then continues to tear fiber by fiber until the final rupture (Fig. 9)

*Ex vivo* mechanical characterization demonstrated that the primary advantage of the polypropylene monofilament meshes was a low modulus of elasticity, a property that may be exploited to enhance mechanical load transfer from biomaterials to the relatively brittle surrounding tissues, thus reducing the incidence of loosening, hernia recurrence, and local wound complications. Mechanical stability is based above all on correct technical application, the quality of the anchoring zone, and the mechanical properties of the mesh itself [11]. The viscoelastic response is an important feature of natural materials and many synthetic biomaterials. This response modifies the stress magnitudes sustained in the restored unit, especially when polymeric chains or networks are a major constituent of the overall structure [12]. A complex combination of cross-linking and collagen type determines tissue mechanical properties.

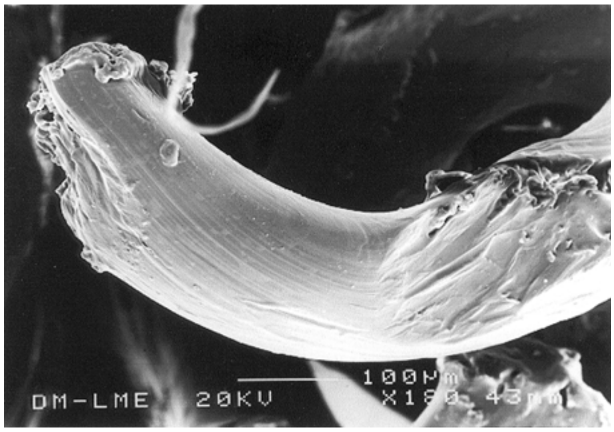


*Figure 9* Final rupture of the polypropylene monofilament.

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