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Polymers in hernia repair – common polyester vs. polypropylene surgical meshes*

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Within the last years meshes have become essential for the repair of abdominal wall hernias. While the type of mesh obviously influences the clinical result, the selection of the best suitable mesh-modification should have favourable effects onto the rate of complications. Available surgical meshes mainly differ in the type and amount of the basic polymers. The most common meshes are made either out of monofilament polypropylene (PP) or multifilament polyester (PET). In the following contribution we studied the functional and histological results of standard and commercially available surgical meshes: a standard heavyweight, large pore-sized PP-mesh (Prolene®), a heavyweight, large pore-sized PET-mesh (Parietex®, coated with bovine collagen) and a low weight small pore-sized PET-mesh (Mersilene®) in a standardised rat model. The meshes are studied by three dimensional stereography, tensiometry, light-(LM) and transmission electron microscopy (TEM), as well as morphometry over implantation intervals of 3, 7, 14, 21 and 90 days. The results proved marked differences between the tested meshes in regard to textile properties, the mechanical function (tensile strength, abdominal wall mobility), as well as the histologically proved tissue reaction. Both heavyweight meshes (PP and PET) revealed an enormous and most similar strength whereas the low weight PET-mesh primarily showed a considerable increase of flexibility. Despite their different structures and their diverse histological response all tested meshes led to a similar and significant reduction of the abdominal wall flexibility. However, the local tissue response of the interface mesh/recipient tissues revealed a significant reduction of the acute inflammatory activity and a significant decrease of connective tissue formation in the case of the low weight PET-mesh Mersilene® compared to both heavyweight mesh-modifications. Mersilene® showed an excellent and relatively inert tissue reaction of the interface compared to Prolene® and Parietex®. Modifications of the mesh-structure (e.g. larger pores) should improve the functional results, in particular, abdominal wall flexibility. However, the use of PET in hernia surgery is at least questionable in respect to the obligate long-term degradation of this polymer.

1. Introduction

The use of polymer meshes in hernia surgery has become essential, particularly in the therapy of recurring or large incisional hernias. They usually consist of non-absorbable polymers like expanded polytetrafluorethylene (ePTFE, e.g. Gore-Tex®), polyester (PET; poly-(ethylene terephthalate), such as Mersilene®, Parietex®) or most often polypropylene (PP; e.g. Marlex®, Prolene®, SurgiPro®, Vypro®).

Common complications of meshes are local wound disturbances including seromas in 30–50% [1–3], discomfort and restriction of abdominal wall mobility in

* This study was performed in adherence to the NIH guidelines for the use of experimental animals and to the guidelines of the "Deutsche Tierschutzgesetz".
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25–50% of all cases [4–6], migration and fistula formation in up to 16% [7–11] and recurrences in up to 34% [7]. In regard to the limited experiences of a sufficiently long follow-up the influence of the mesh material to the rate of complications is discussed controversially. Whereas *Leber* advised not to use PET-meshes for their significantly increased complication rate *Morris-Stiff* found no considerable effect of the mesh material in a review of the outcome of non-absorbable meshes within the abdominal cavity [7, 12]. Anyhow, to avoid mesh-related complications we have to focus on both the implantation technique and the mesh characteristics [13].

The selection of the best suitable mesh-modification should be based on the textile properties and the induced tissue reaction with its functional consequences within the host. Variations of either the textile-structure or the type of used polymer are assumed to affect the biocompatibility of the mesh.

PP-meshes such as Prolene® are heavyweight meshes made of monofilaments, therefore causing a considerable stiffness and developing sharp edges after cutting. In contrast, meshes of PET are constructed of multifilaments, whereas, Parietex® achieves material properties similar to that of the heavyweight PP-mesh Prolene®.

Recent studies of PP-meshes illuminated the marked effect of the mesh-weight and surface area on the inflammatory response of the host tissue [14]. Therefore, the aim of the present study was to analyse the effect of the polymer, either PP or PET, in regard to different mesh weights (heavyweight PP Prolene®, low weight PET Mersilene® and heavyweight PET Parietex® mesh). All meshes have been tested regarding their capability to reconstruct abdominal wall defects in rats. Before implantation the meshes have been investigated in a textile analysis to evaluate the material properties and the mechanical features of each mesh-modification. Furthermore, we have determined the abdominal wall mobility, the tensile strength as well as the local histological and ultrastructural tissue reaction for 3 to 90 days after implantation.

2. Materials and methods

2.1. Experimental animals

180 male Wistar rats (250–300 g) were studied. The animals were housed under conditions of constant light and temperature and received a complete diet of rat feed ad libidum throughout the entire study, which was performed according to the rules of the "Deutsche Tierschutzgesetz" (AZ 23.203.2 AC 18, 17/94) and to the NIH guidelines for the use of laboratory animals. The animals were randomly divided into three groups (each $n = 45$). As controls served a sham-operated group, meanwhile the three test groups obtained the PP and PE mesh-modifications Prolene®, Mersilene® and Parietex®. Biomechanical observations were carried out on 25 animals (each mesh $n = 5$/day), morphology and morphometry are performed on 20 animals of each group (each mesh $n = 4$/day).

2.2. Mesh modifications

In the present study we compared the heavyweight Prolene® (Ethicon®; Germany), the low weight Mersilene® (Ethicon®; Germany) and the heavyweight Parietex® (Cogent®; France) meshes (Table I). The textile analysis to estimate the mechanical material properties was done according to existing DIN rules or EN/ISO-classifications (DIN 53884).

TABLE I Material properties of the used meshes

Name	Prolene®	Mersilene®	Parietex®
Material	PP	PET	PET, coated with bovine collagen
Type of filament	Duofilament	multifilament	multifilament
Weight (g/m^2)	108.5	39.5	129.6
Proportion of pores (%)	84	90	79
Maximum pullling force (N/50 mm)			
Vertical	597	205	391
Horizontal	767	100	636
Bending stiffness N/m^2			
Vertical	6700	400	1000
Horizontal	1290	100	2420
Subsequent tearing force (N)			
Vertical	0.0	6.4	33.6
Horizontal	4.4	6.8	27.8
Tearing out the seam (N)			
Vertical	57.0	15.2	68.5
Horizontal	74.6	15.5	55.4
Crease recovery angle, 30 min.(°)			
Vertical	125	87	73
Horizontal	135	127	155
Testing the pressing through the stamp			
Maximum force (N)	2369	443	2026
Tensile strength per mm (N/mm)	9.0	1.95	9.04
Elongation at maximum force (%)	44	37	36
Elongation at 1.6 N/mm (%)	7	7	4

To test the force of pressing through the stamp (modified DIN 54307) a circular mesh sample with a radius of $r = 56,4$ mm and with a test area of 0.01 m^2 was clamped. The fixed mesh is finally loaded with a spherical stamp (radius $r = 50$ mm; velocity $V = 50$ mm/min) until rupture occurs. Based on the forces and the resulting stretching we calculated the circumference where the stamp lost contact with the mesh. The force leading to the rupture of the mesh is divided by the corresponding circumference to calculate comparable forces per 10 mm (N/10mm). The deformation (%) corresponds to the increased mesh area compared with the initial area of the mesh before deformation. In regard to the maximum physiological tensile strength of 16 N/10 mm [15–17] we calculated the elasticity of the mesh at a strength of 16 N/10 mm during the testing of pressing through the stamp. The tensile strength of the whole mesh is estimated according to DIN 53857 using a tensiometer with a sample width of 50 mm, taken in a distance of 100 mm from the margin of the mesh. The bending stiffness is measured with a cantilever-device at 10 mesh-samples of each mesh-modification (N/m^2, DIN 53882) and defined by the resistance of the mesh to bending-forces (the mesh-weight). All tests are performed in longitudinal (horizontal) and cross (vertical) direction of each mesh.

2.3. Surgical procedure

Anaesthesia was achieved with a mixture of ketamine hydrochloride (80 mg/kg intraperitoneal (i.p.)) and xylazine (8 mg/kg i.p.). The skin was shaved and disinfected with Betaisodonna® solution. After midline incision, a full-thickness defect was performed resecting the rectus muscles with peritoneum (except skin), 20 mm distal of the xiphoid, on an area of 20×30 mm *en bloc*. Each mesh (20×30 mm) was fixed as real abdominal wall replacement continuously in inlay position with 5/0 Prolene® without overlap between muscles and prosthesis. Skin closure was finally obtained with 3/0 silk continuous sutures. No antibiotic treatment was given before or during the experiments. Sham controls were performed by simple closure of the midline laparotomy with continuous 5/0 Prolene®.

2.4. Observation periods

From each group ($n = 45$) nine animals were killed at day 3, 7, 14, 21 and 90 after implantation of the meshes. Throughout the whole observation period of 90 days all animals of each group were objectively controlled daily for local and systemic (wound)-complications.

2.5. 3D-photogrammetry

After sacrificing the animals each time the bending stiffness of the abdominal wall was determined using a videographic method based on three dimensional photogrammetry [18]. For measurements of the abdominal wall mobility the abdominal cavity was filled with physiologic saline solution under controlled and increasing pressure levels (0–9.3 kPa). Simultaneously, a square pattern was projected onto the surface of the abdominal wall covering an area of 40 mm $\times$ 40 mm. The deformation of the projected squares was documented by a video system. Pictures obtained from this system at different intraabdominal pressure levels were digitised and finally analysed by a special computer system. The curvature of the mid-region was calculated from the deformation of the squares by an automatic pattern recognition program (Institute of Aerodynamics, RWTH-Aachen, Germany [18].

The videopictures were digitalized and analysed by a Windows®-software based on Matlab® version 4.2b. The relative position of the object to the camera level is defined by a standard cube. The area of interest is manually determined by points at the edge of the marking on the abdominal surface. All pictures were defined with a picture processing algorithm by a co-ordinate transformation based on the position of these points and the relative position of the low grid junctions. The coordinates of the abdominal surface are calculated and presented three dimensionally with the help of standard values and trigonometrical equations. The surface is recalculated as a polynom of fourth grade, the curvature as its second derivation (m^{-1}). For comparison we always determined the mean horizontal curvature of the centre (middle 50% of all values).

2.6. Tensiometry

After determination of the abdominal wall mobility the tensile strength was measured using a tensiometer. Measurements were made on 20 mm wide strips obtained after excision of the whole mesh with the surrounding abdominal wall. The tests were carried out directly after explantation of each mesh. The tensile strength of the suture zone was determined firstly, then that of the mesh alone (velocity of stretching 10 mm/min).

2.7. Morphological study

Specimens were studied by light (LM)- and transmission electron microscopy (TEM). For LM tissue specimens were fixed in a buffered 10% formalinbath, embedded in paraffin, and sections were stained with haematoxylin and eosin (H&E), as well as periodic-acid Schiff (PAS) plus diastase and elastica van Gieson (EvG). For TEM tissue specimens were fixed in 3% cacodylate-buffered glutaraldehyde for 30 hours. After fixation in osmium, buffering in 0.1 M cacodylate, they were dehydrated in ethanol and embedded in Epon. Semi-thin sections (1.0 μm) were stained with methylene blue-azure II. Ultra-thin sections were mounted on copper grids, stained with uranyl acetate and lead citrate and examined with a transmission electron microscope (Philips, EM400T).

2.8. Morphometry

The morphometric evaluation consisted of a quantitative cell analysis of the inflammatory reaction and the soft-tissue reaction. Morphometry was performed in the centre and the suture zone of the mesh. Cells in 10 fields of 5 HE slides at a grid of 10 points (140 $\times$, area 0.1 mm^2) and within the interface of 0–300 μm (area 636 μm^2) on the TEM were counted. Parameters measured are the inflammatory infiltrate (partial volume (PV) %), connective tissue (PV %), macrophages (%), lymphocytes (%), granulocytes (%), giant cells (%), and fibroblasts (%).

2.9. Statistics

Statistical analysis is carried out using Statistical Package for Social Sciences (SPSS) - software. All functional and morphological results were analysed for statistical significance using a corrected analysis of variance [LSD-test (Least-Significant-Differences] according to Bonferroni), followed by independent *t*-tests in cases of significant differences. *P*-values <0.05 were considered to be significant. The data are given as mean $\pm$ SD throughout the entire study.

3. Results

3.1. Material properties of the native meshes (Table I)

The heavyweight PP-mesh Prolene® consists of two monofilaments with a gauge of 21 tex and a weight of 109 g/m^2. It possesses an enormous tensile strength both for maximum pulling forces at stripes of 50 mm and in the test pressing through the stamp. In contrast, the subsequent tearing force is considerably low particularly in vertical direction, indicating a certain asymmetry of the mesh structure. Due to a large pore size of 1,6 mm and despite the thick filaments the partial

volume of the pores reaches 83%. The bending stiffness varies between 1300 N/m^2 in horizontal direction and 700 N/m^2 vertically. The crease recovery angle after 30 min as a parameter of memory properties are considerably high with 125° vertically and 135° horizontally. The high forces necessary to tear out the seam confirm the high strength of this mesh-modification.

The low weight mesh Mersilene® consists of thin PET multifilaments (6 tex) with a comparatively low weight of 40 g/m^2 (37% of Prolene®, 30% of Parietex®). With a diameter of maximum 1 mm it has a small porous structure. Nevertheless, as a consequence of its structure the proportion of pores reaches 90% of the whole mesh-surface. The bending stiffness is very low (factor 25 vertically and factor 240 horizontally compared with Parietex®). The crease recovery angle after 30 min vertically reaches 87° and horizontally 127°. The test of tearing out the seam reaches 1.5 N/mm, almost symmetrically in horizontal and vertical direction. Compared with the heavyweight meshes it is more flexible. The subsequent tearing force is considerably less than for the heavyweight mesh Parietex® but exceeds the force of the PP-mesh. The tensile strength of a mesh-strip of 50 mm width exceeds 200 N in vertical direction, but in horizontal direction the tensile strength reaches only 100 N. Testing the pressing through the stamp the tensile strength of 1.95 N/mm is calculated at an elasticity of 7% at 16 N.

The heavyweight mesh Parietex® consists of thick bundles of bovine collagen-coated PET multifilaments fibres (98 tex) and an absolute weight of 130 g/m^2. Its tensile strength exceeds the 4–5 fold of the Mersilene® mesh and appeared similar to that of the heavyweight PP-mesh. Although, the pore-diameter is about 1.5 mm the overall proportion of pores again is reduced to 79% due to the thick polymer fibres. The bending stiffness is high both in vertical and horizontal direction. The crease recovery angle after 30 min reaches vertically 73° and horizontally 155°. The test of tearing out the seam exceeds 5.5 N/mm, almost the same in horizontal and vertical direction indicating the high symmetry of the mesh. The subsequent tearing force is very high with 34 N vertically and 28 N horizontally. The tensile strength of a 50 mm wide mesh-strip in vertical direction exceeds 390 N, and in horizontal direction even 636 N. Testing the pressing through the stamp the tensile strength is calculated to 9.04 N/mm at an elasticity of 4% at 16 N.

3.2. Macroscopic observations (Table II)

The implantation of three meshmodifications led to specific local and systemic wound-complications. Following the implantation of the low weight PET-mesh we observed seromas in 25 of 45 cases (56%). With both PET-meshes we macroscopically saw a significant increase of the rate of local inflammation compared with the sham-operated group or the PP-group, respectively. Generally, the local inflammatory reaction occurred without any suppuration but sometimes with wound edge separation, particularly, in the case of the heavyweight PET-mesh Parietex®. In two cases of the low weight PET-mesh Mersilene® we observed a visible protrusion after 90 days, so that the stereography could not be performed correctly. The explantation revealed a capsule formation around the mesh, filled with water at increasing levels of the intraabdominal pressure. During the whole observation period four rats died without any obvious relation to the meshes.

TABLE II Macroscopic observations after implantation of the low weight PET-mesh and the heavyweight PET- and PP-meshes compared with the sham operated group

Complications	Control	Heavyweight PP	Low weight PET	Heavyweight PET
Seroma	-	2	25*	4
Hematoma	-	1	3	5
Local wound inflammation	1	5	16*	21*
Protrusion	-	-	2	-
Deaths	-	3	1	3

$n = 45$ each group; * = $p < 0.05$ versus control.

3.3. Curvature measured by three dimensional stereography (3D-photogrammetry; Table III; Fig. 1A–C)

The calculated curvature of the abdominal wall of the sham operated, control rats showed a slow decrease at low pressures, whereas parallel to the rising pressure

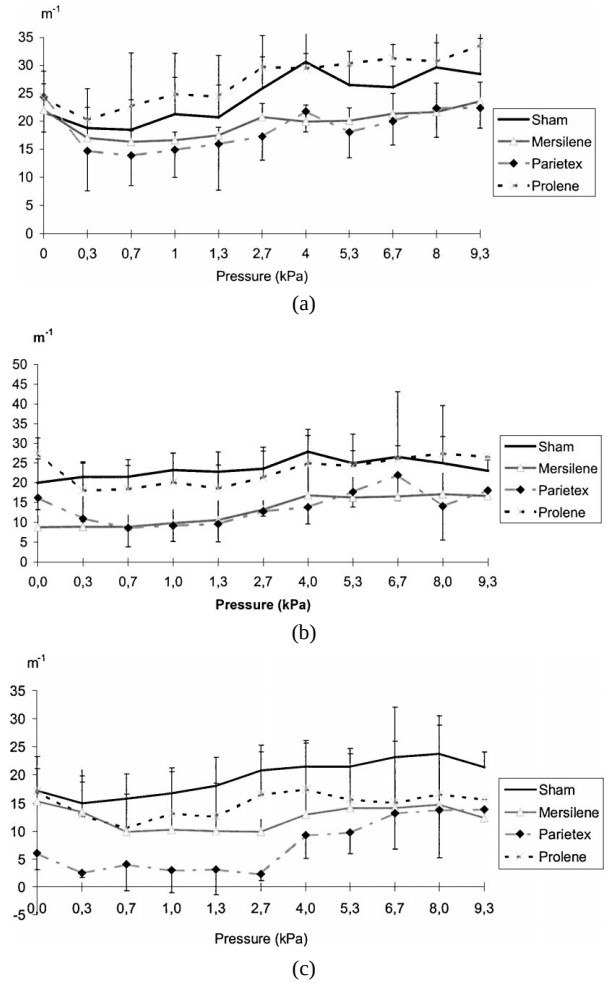

Figure 1 Mean (+SD) curvature (m^{-1}) measured by 3D-stereography, (A) 14 days, (B) 21 days and (C) 90 days ($n = 5$) after implantation.

TABLE III Abdominal wall mobility by three dimensional stereography: Mean values of curvature (m^{-1})(SD) from 0 to 9.3 kPa ($n = 5$).

Curvature (day)	3	7	14	21	90
Control	24.1 (6.9)	24.0 (9.1)	23.9 (6.7)	23.5 (6.5)	19.3 (5.1)
PP	27.5 (6.9)	29.1 (5.3)	27.1 (9.1)	22.5 (6.6)	15.1 (2.5)
PET					
Low weight	14.8 (6.2)*#	25.8 (8.8)	20.2 (4.3)#	13.0 (5.8)*#	12.6 (4.4)*
Heavyweight	14.6 (7.0)*#	25.5 (5.7)	18.4 (5.8)#	13.5 (5.7)*#	7.2 (6.3)*#

PE * = $p < 0.05$ versus control; # = $p < 0.05$ versus PP.

the curvature correspondingly increased to values between 25 and 30 (m^{-1}). After 90 days the curvature was constantly reduced as a consequence of the growth of the rats.

The implantation of a heavyweight PP-mesh did not cause an obvious alteration of the abdominal wall mobility within the first three weeks, whereas after 3 months the PP-mesh revealed an increase of stiffness of the mesh area.

In contrast, both PET-meshes led to an increase of stiffness already after 14 and 21 days. Though the low weight PET Mersilene® mesh had been considerably more flexible in its textile form, it also revealed a markedly reduced curvature of about 10 to 15 (m^{-1}) ($p < 0.05$ after 21 and 90 days). The heavyweight PET-mesh Parietex® developed the most extended stiffness of all tested materials, starting already 14 days after implantation.

At the end of the observation period all implanted materials led to a fairly similar reduction of the abdominal wall mobility with a rising level from heavyweight PP, low weight PET and heavyweight PET.

3.4. Tensiometry (Table IV)

After sham operation the tensile strength of the suture zone was reduced to 0.34 N/mm after 3 days, then increased constantly, finally reaching the full strength of 1.6 N/mm at the end of the observation period of 90 days. In comparison the tensile strength of the suture zone after implantation with the various meshes indicated no marked differences to the controls. The tensile strength of the explanted mesh samples remained almost constant except for the low weight Mersilene® mesh whose initially very low strength doubled from initially 2.0 N/mm to 4.0 N/mm after 3 months.

TABLE IV Mean tensiometric results (SD) of the suture zone (A), and of the mesh (B) (N/mm)($n = 3$, horizontally)

A

Day	Sham	Heavyweight PP	Low weight PET	Heavyweight PET
3	0.3 (0.2)	0.2 (0.0)	0.4 (0.0)	0.2 (0.0)
7	0.8 (0.2)	0.8 (0.2)	1.1 (0.1)	0.9 (0.1)
14	0.8 (0.2)	0.8 (0.4)	1.2 (0.2)	1.0 (0.2)
21	1.1 (0.4)	1.0 (0.3)	1.4 (0.2)	1.2 (0.4)
90	1.7 (0.2)	2.2 (0.7)	1.8 (0.0)	1.4 (0.2)

B

Day	Heavyweight PP	Low weight PET	Heavyweight PET
3	8.8 (1.2)	2.0 (0.2)	8.5 (0.0)
7	8.4 (0.9)	2.2 (0.3)	7.0 (0.9)
14	8.4 (1.1)	3.0 (0.4)	4.0 (1.3)
21	8.6 (0.6)	4.1 (0.6)	6.2 (0.8)
90	8.0 (1.3)	4.0 (0.8)	8.4 (2.8)

3.5. Morphological analysis (Fig. 2A–D)

The implantation of a heavyweight PP-mesh Prolene® usually initiated a pronounced acute inflammation throughout the whole mesh area. A moderate, purely serous oedema had vanished after 2 weeks, being replaced after one week by fibrosis with dense infiltrates of polymorphonuclear granulocytes (PMN) and macrophages. However, whereas PMNs showed a descending density in the interface, macrophages revealed a rising colonisation rate and were the dominating inflammatory cell type after 90 days. Moreover, in TEM of the contact zone of the PP-fibres to the mesh polymer macrophages were demonstrated to transform to epitheloid like cells, and, in parallel, with a continuously multiplying number of multinucleated giant cells. Interestingly, Prolene® displayed at day 21 a remarkable decline in the number of macrophages and epitheloid cells in the interface in contrast to the count of PMNs. An enlargement of the amount of apoptotic macrophages and epitheloid cells, respectively, could be observed by TEM. Overall, at the end of the observation period a moderate persisting inflammatory interface response could be evaluated in this rodent model. The inflammatory reaction at day 90 should be interpreted as a still active inflammation with persisting PMNs and a major composite, epitheloid granulomas from the foreign body type with a minor or moderate amount of multinucleated giant cells.

The cell infiltrate directly in contact with meshes of the low weight PET-type (Mersilene®) mainly consisted of macrophages, eosinophils and lymphocytes. Plasma cells were constantly rare as well as polymorphous granulocytes (PMNs). The latter were only observed at days 3 and 7 as consequence of the operation. A peculiarity of the low weight Mersilene® mesh was the marked fibrin deposition at days 3 to 14 which was observed in this intensity only in this PET-mesh modification. On the other hand the histological reaction of the Mersilene® mesh was relatively uniform with a marked accumulation of macrophages at days 3 to 14. After day 14 the macrophages matured to epitheloid cells and merged to typical foreignbody giant cells. At the end of the observation period of 90 days histology revealed a chronic inflammatory reaction from the foreign-body type without acute inflammatory cells, in particular, without PMNs. Collagen deposition

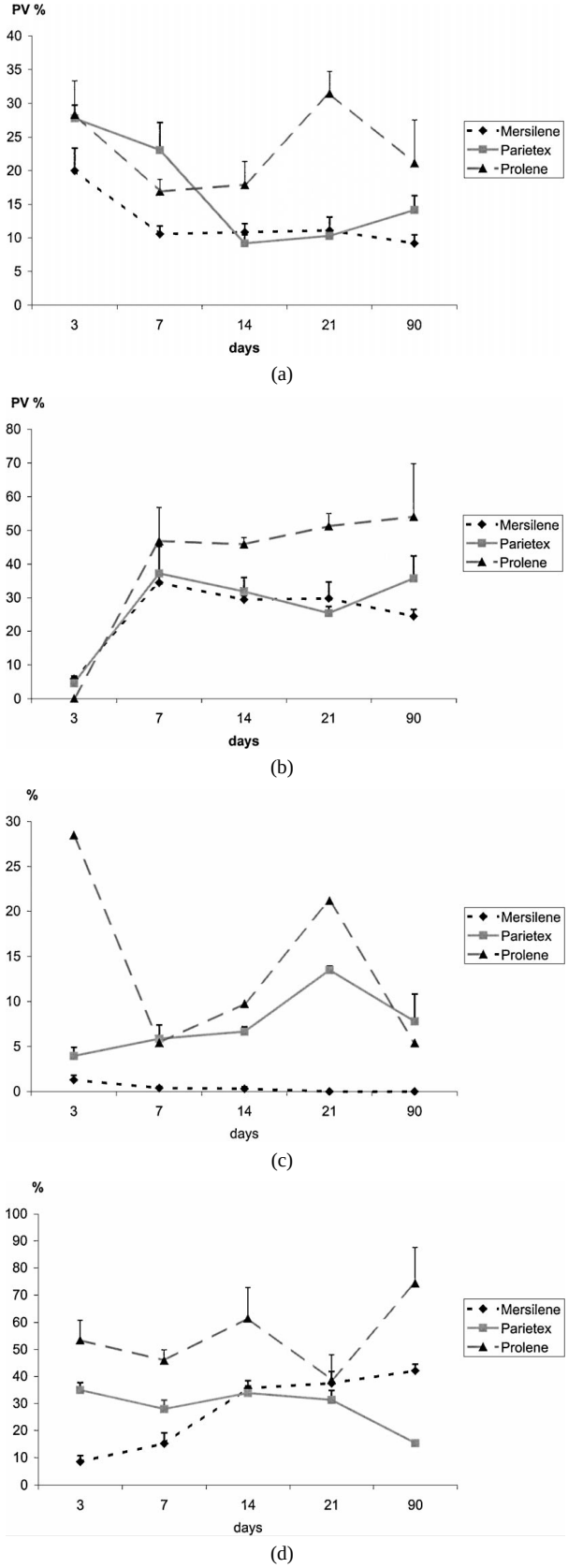

Figure 2 Morphometric data (mean + SD) of the partial volume (%) of (A) the inflammatory infiltrates and (B) the connective tissue as well as the percentage (%) of (C) granulocytes and (D) macrophages at the total cell count.

altogether was low. After 90 days the Mersilene® mesh was integrated into a well defined smooth tissue capsule surrounding each multifilament fibre-unit.

The heavyweight PET-mesh Parietex® revealed a contrary histological and ultrastructural picture during the observation period of 90 days compared to Mersilene®. Basically, the inflammatory reaction was characterised by the formation of granulomas from the foreign-body type, however, in size and quantity less pronounced. Moreover, the recipient tissues were loosed up by a marked oedema of the interstice with a maximum at day 21 and still clearly detectable after 90 days. The most important difference, however, was the finding that the interface showed an acute inflammatory reaction characterised by the evidence of PMNs and areas of fibrinoid necrosis. At day 21 the inflammatory infiltrate mainly consisted of PMNs focally forming micro-abscesses and at the end of the experiments at day 90 PMNs were still the leading cell-group. Corresponding to the significant increase of PMNs cell-groups indicating a chronic inflammation such as lymphocytes, giant cells and macrophages or transformed epitheloid cells, respectively, were usually decreased compared to the low weight variant Mersilene®. Moreover, the number of fibroblasts and the amount of deposited collagen-fibres as well as the rate of vascularisation were significantly decreased after 90 days.

4. Discussion

The task of polymer-meshes in hernia surgery is the reinforcement of the abdominal wall, at best without any restriction of abdominal wall mobility and without any inflammation. Surgical meshes are thought to strengthen the abdominal wall by induction of an intense inflammatory reaction resulting in a strong scar plate completely surrounding the mesh-fibres [14, 19]. The mechanical support of the abdominal wall by meshes has been proven in many clinical and experimental studies [20–23].

However, an increasing number of reports demonstrates potential risks of different mesh-variants such as fistulas and erosions of the small and large bowel [11, 24]. Although, contradictory results about a direct effect of the implanted mesh material to the rate of complications some material related complications cannot be excluded. Anyhow, the application of a mesh requires a conscious selection of the mesh-material. Common mesh polymers predominantly used for hernia repair are based either on PP or on PET. Successful clinical reports are available for all meshes investigated in this study [25–27]. However, because some aspects of the tissue response are supposed to become obvious with a latency of several years, the awareness of the textile properties and the knowledge of the principally induced tissue reaction might be helpful for the decision of the best suitable mesh.

Apart from the modification of the basic polymer another major difference of surgical meshes is the amout of manufactured material with different mesh-weights and -surface areas. The present study reveals that the type of polymer, the weight in g/m^2, the proportion of pores in % and, therefore, the surface area in contact with the recipient tissues play key roles for the evaluation of biocompatibility and the host reaction.

The original stiffness and the intense scar formation after implantation of the PP-meshes Marlex® and Prolene®, respectively, are supposed to be responsible

for abdominal wall restriction [19, 28, 29]. The textile analysis as the essential base of mesh research confirmed the inappropriate strength of the heavyweight PP-mesh Prolene® (>9.0 N/mm) [29] compared to the physiological required tensile strength of less than 1.6 N/mm [15, 30]. Furthermore, the interaction with the ingrown scar resulted in a marked decrease of abdominal wall compliance, starting after 3 weeks parallel to the histological development of connective tissue. Interestingly, the ingrowth of scar tissue did not significantly increase the tensile strength of the meshes.

In contrast to the monofilament meshes of PP with their marked stiffness, meshes made of PET-multifilaments usually are smooth and flexible. Interestingly, after implantation of the more elastic low weight PET-mesh Mersilene® the following decrease of the abdominal wall mobility was even more pronounced indicating the importance of the mesh-construction for the tissue response, the *in vivo* function and the mesh integration into the artificial abdominal wall after implantation.

The measurement of the tensile strength exhibits that the suture zone is the mechanically weakest area after mesh-implantation. These findings confirm the unsuitability of surgical meshes to reinforce the abdominal wall in inlay-position and explain the high recurrence rates for this technique [31–33]. Moreover, the data prove from the mechanical point of view that surgical meshes in general have only to be a little bit stronger than the anchoring zone connecting the mesh-structure with the recipient tissues.

The examination of the tissue response and in particular the measurement of the partial volumes of cells reveal marked differences between the two polymers. Furthermore, the morphometry of the cells allows a more detailed graduation of the tissue response and can reflect differences even between meshes made of the identical polymer. Thus, whereas the inflammatory reaction to PET seems to be less than for PP, the use of a large amount of PET can stimulate the inflammation as seen for PP. As recently reported by our group [14] the weight and structure of various PP-meshes considerably influence the extent of inflammation and, furthermore, the integration of mesh fibres into the recipient tissues [3, 34]. The present study reports identical observations for the group of PET-meshes. The low weight PET-mesh Mersilene® causes a pure chronic inflammatory response, whereas, the heavyweight variant Parietex® reveals an acute inflammatory reaction, although, this peculiar mesh-modification is coated with bovine collagen. However, the surface modification seems only to exhibit a protection for the first 14 days in our *in vivo* model with a resorption of the bovine collagens and a significant amplification of the acute inflammatory reaction after 21 days.

Interestingly, PET causes a pronounced fibrinous seroma after 3 days, similar to the frequently seen seromas in humans. Furthermore, Tang [35] proved in a mouse-model the importance of fibrin for the regulation of the inflammatory response probably influenced additionally by the interaction with the phagocyte-integrin MAC-1 (CD11b/CD18). The tissue reaction to the heavyweight PET-mesh on the one hand showed low numbers of lymphocytes and macrophages compared to the low weight PET-mesh Mersilene® and on the other hand elevated counts of PMNs as found after implantation of heavyweight PP-meshes. In general, the pronounced induction of an acute inflammation by Parietex® resembles more the heavy-weight PP-meshes than the low weight PET-mesh Mersilene®. Recently, it could be shown that the inflammatory reaction of PET can be decreased by a surface-modification or-coating with a fluoropolymer [36].

The particular analysis of the measured partial volumes after 90 days exhibits a strong correlation of the extent of inflammation to the amount of connective tissue. This underlines that the formation of the collagen tissue and its extent depends on the degree and activity of inflammation, although, none of these values correlate directly to the mobility of the abdominal wall in an outstanding way.

Fibroblasts appeared to be well representative of the connective tissue except for the heavyweight PET-mesh Parietex®. This mesh showed a constant decrease after the second week of implantation, probably due to the persistent acute inflammation hindering the development and maturing of the scar tissue. Newer results of our group indicate significantly elevated rates of Ki-67- and TUNEL-positive fibroblasts in the interface of the Parietex® mesh compared to other mesh-modifications proving a permanent cell turn over in the fibrous capsule with increased cell proliferation- and cell death-rates (data not shown) and indicating a high rate of scar tissue remodelling. This phenomenon might explain the lowered content of connective tissue compared to other investigated heavyweight mesh-modifications.

Although, it is well known that PET is hydrolytically degraded after long-term implantation [37–39] we did not expect any fragmentation *in vivo* after 90 days. In 1997, Riepe *et al.* found that in vascular grafts of Dacron® the hydrolytic degradation takes 10 years to decrease the bursting strength on a level of 31% and 25–39 years for complete fragmentation [40]. Meanwhile there are first reports of clinically detected mesh degradation [25, 41, 42]. In contrast to the inevitable degradation of PET the permanent PP is assumed to preserve its chemical and mechanical integrity for years, and for this reason seems to be advantageous for hernia surgery with implantation intervals of years or even decades.

Recent animal experiments could demonstrate the improved integration of low weight, large pore sized PP-meshes into the recipient abdominal wall in regard to the tissue response as well as the mechanical properties [17]. These mesh modifications with a reduction of PP to less than 25% of the heavyweight PP-mesh Prolene®indicated a chronic inflammatory reaction at the interface quite similar to the one of the low weight PET-mesh Mersilene®. Furthermore, in contrast to the small pores sized meshes they preserve an unlimited abdominal wall mobility by avoiding the formation of a scar plate embedding the complete mesh [17].

In conclusion, the presented differences of tissue reactions as well as the functional consequences due to implanted meshes confirm the necessity of animal

models in hernia research. Although, results of rodent animal models cannot be directly compared to the human situation the obvious advantage of animal experiments is to get a better evaluation and understanding of the *in vivo* function of implanted meshes under standardised experimental conditions. In regard to the frequently formulated reservation to animal experiments in rodents, we can confirm fairly similar results in pigs, dogs and rabbits. Meanwhile we have even found the same characteristic tissue response in humans at explanted meshes, which have been incorporated for several years [42].

The test of polymer-meshes for abdominal wall repair in future should include functional tests like 3D-stereography and tensiometry. Both methods give a close insight into the mechanical properties of each mesh modification before and after implantation. The combination of the functional test-results with morphological data of the tissue response allows a concrete pre-clinical evaluation of new mesh-modifications.

In regard to the initially extended inflammation of heavyweight PP-mesh modifications the use of PET-meshes seems to be favourable, at least if the amount of incorporated polymer is low (e.g. Mersilene®). A disadvantage of PET is the early tendency to form seromas, the construction out of multifilaments in the case of contaminated wounds and, in particular, the lack of long-term stability, which makes its use for a durable closure of hernias questionable.

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