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Perspective article

Surface science in hernioplasty: The role of plasma treatments

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ABSTRACT

The aim of this review is to clarify the importance of surface modifications induced in biomaterials for hernia-repair application. Starting from the pioneering experiences involving proto-materials as ancient prosthesis, a historical excursus between the biomaterials used in hernioplasty was realized. Subsequently, after the revolutionary discovery of stereoregular polymerization followed by the PP application in the biomedical field performed by the surgeon F. Usher, a comparative study on different hernia-repair meshes available was realized in order to better understand all the outstanding problems and possible future developments. Furthermore, since many unsolved problems on prosthetic devices implantation are linked to phenomena occurring at the interface between the biomaterials surface and the body fluids, the importance of surface science in hernioplasty was highlighted and case studies of new surface-modified generations of prosthesis presented. The results discussed in the following evidence how the surface study are becoming increasingly important for a proper knowledge of issues related to the interaction between the living matter and the artificial prostheses.

1. Introduction

The term "hernia" indicates an abnormal protrusion of an organ or its part outside the body cavity which normally contains it. Several kinds of hernias are known. According to the visceral dislocation, hernias can be classified as internal (inside the body, not-evident) or external (evident from outside). The most frequent ones occur in the abdomen (epigastric, umbelical, ventral, inguinal, and femoral hernia) and all of them required a surgical intervention [1–4]. The placement of a synthetic mesh is the standard technique to close mechanically the hernia cavity and reinforce the body wall previously damaged by facilitating the cellular tissue growth [5,6].

The most commonly used synthetic meshes for hernioplasty are made in polypropylene (PP) with controlled fiber dimensions and sizes. These polymeric devices have been widely applied because of PP properties (i.e., physical and chemical inertness, stable, non-toxic, non-immunogenic, flexible, easily integrated by surrounding tissues, and so on) [7]. Unfortunately, complications and risks during this routinely surgical intervention and in the post-operative phase are still present [8,9]. Among all the clinical complications which patients may suffer after a hernia-repair surgical operation, the appearance of inflammatory response [10], which in extreme cases can induce even the prosthesis rejection [11] is the most frequent. Inflammations can be due to bacterial infections generated as a result of the prosthetic device application [12]. In fact, when the pores in the surgical meshes are smaller than 10 μm, bacterial cells (typically *Staphylococcus aureus*) with an average size of 1 μm can easily permeate the surgical mesh forming biofilms (see for instance Fig. 1) [13–15]. A biofilm is an aggregate of microorganisms in which cells, which are frequently embedded within a self-produced matrix of Extracellular Polymeric Substance (or EPS, mostly composed by nucleic acids, proteins and polysaccharides), adhere to the prosthesis surface [16]. On the other hand, macrophages (the cells of our body whose function is to phagocytate the bacteria) have a higher diameter comparing to the pores of surgical meshes, thus they cannot prevent the bacteria infiltration. This sterical difference may be the potential cause of the proliferation of bacteria and, consequently, of infections. Additionally, macrophages can also accumulate nearby the synthetic (and contaminated) implants (histiocytosis), contributing to the implant rejection as in the case of foreign-body reactions [17,18].

In order to avoid all these undesired phenomena related to bacteria proliferation, macroporous implants, able to promote the direct contact between macrophages and bacteria, were proposed [5,19]. The adhesion of bacteria at the surface of biomedical devices is a key-point in the infection development. Unfortunately, the

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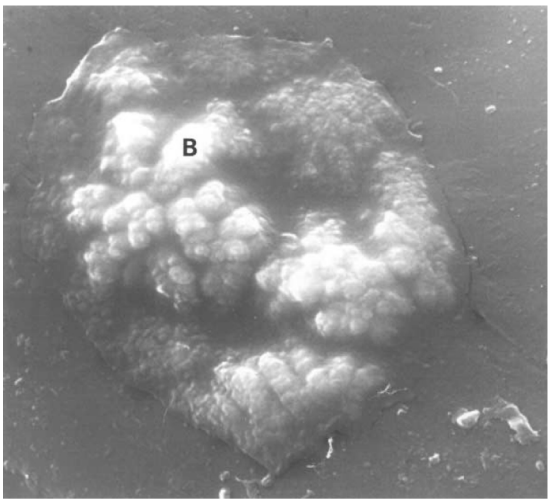

Fig. 1. SEM micrograph illustrating the formation of a typical biofilm (B) on a polypropylene filament. (Magnification 300x). Reprinted with permissions from [15].

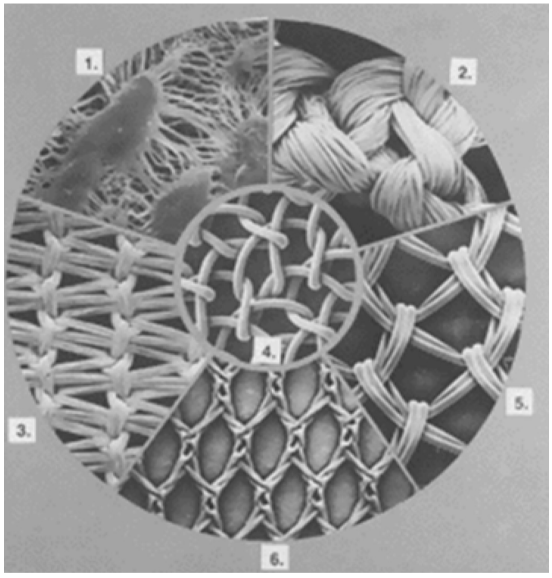

Fig. 2. SEM micrographs of: 1) PTFE soft tissue patch (ePTFE or Gore-Tex); 2) multifilament PTFE mesh (Teflon); 3) multifilament PP mesh; 4) monofilament PP mesh; 5) double filament PP mesh; 6) multifilament polyester mesh (or PET). Reprinted with permissions from [29].

mechanism of microorganisms' adhesion involves many aspects which are still unclear. Hence, the local prevention of bacterial colonization together with the action of antibiotics directly *in situ* is becoming mandatory procedures in the hernioplasty treatment. Biomaterials surface modification has emerged as a possible route to follow in order to overcome bacteria proliferation and adhesion at the polymeric surface.

Aim of this work is to evidence the importance of surface science in hernia-repair surgery. To do this, a brief historical excursus on hernia-repair biomaterials is proposed, focusing on the evolution of biomaterials: based on bulk properties evaluation (past view) or based on surface modification procedures (new generation of prostheses). Furthermore, possible future scenarios together with the potential evolution of new-generation biomaterials with smart peculiar properties at the surface level are discussed.

2. From metals to polymers: the polypropylene tale

The IUPAC definition of a biomaterial is a material exploited in contact with biological systems, mostly living tissues, organisms or microorganisms [20]. These materials constitute the prosthetic devices and may be natural or synthetic (typically metals, ceramics, polymers and composites). Since the ideal biomaterial does not exist yet, many studies have been performed to find the best solution in term of biofunctionality and biocompatibility, where the first term refers to the ability of devices of playing a certain physical and mechanical function, whereas the second term is connected to the interactions between prosthetic materials and the living tissues, favoring the biomaterial assimilation and avoiding the prostheses rejection.

Biomaterials and their use in the biomedical field is a path that develops from the very ancient times. In fact, in order to ensure an improvement of the quality of life, several attempts have been conducted to replace defective parts of the human body by using extracorporeal materials. The oldest prosthetic device in the human history was probably the ancient Egyptian false big toe from the Theban necropolis. This rudimentary artificial prosthesis has been dated from around 950–710 years BC and was made mostly in wood and leather, i.e., the only proto-biomaterials available at that time [21].

Obviously, the technology development, together with a better understanding of both biological mechanisms and human body composition, significantly favored the evolution in the biomedical field in terms of new possible biocompatible materials to be used and requirements necessary for improving the properties of artificial extracorporeal devices.

Going back in time to the abdominal surgery history, metals (mainly gold and silver) were the first materials utilized as sutures by ancient Greeks and Romans [22]. Between metals, silver came out as preferred material for wires thanks to its peculiar antibacterial properties, and the German surgeons Goepel and Witzel [23,24] made first silver devices at the end of 1800. Even if these early prostheses presented lots of clinical problems mainly due to metals mechanical properties (for instance high fragility and rigidity), their use went on until 1960 [25]. When the last attempts were made with tantalum gazes and stainless steel meshes [26], but always with negative results, also several exotic materials were tested as possible alternatives to metals, for instance regenerated cellulose fabrics, polyvinyl sponges, silicone sheets and carbon fibers [24].

Finally, after the Second World War, the plastics industry developed and found its economic and technologic boom. Surgeons, sensitive to new materials development for biomedical applications, considered this opportunity and several new materials-made devices with promising characteristics became available.

Nylon was the very first synthetic polymer to become widely available and applicable in hernioplasty [27]. Additionally, the peculiar shape of such new prosthesis (i.e., meshes of knitted or woven fibers) became the prototype for the modern commonly-used hernia-repair devices. Unfortunately, even if the polymeric route seemed to be very promising, these synthetic polyamides presented lots of undesirable troubles, first of all the loss of mechanical properties (tensile strength) when implanted into tissues due to denaturation induced by hydrolysis.

Polyester meshes made by polyethylene terephthalate (PET) under the commercial name of Dacron or Mersilene became available in 1956 [28]. Soon after, other polymers were selected [29], such as polypropylene (PP) and Teflon (polytetrafluoroethylene, PTFE) [30]. The latter, once expanded, generates ePTFE sheets (or Gore-Tex) that was surgically applied for hernia defects care for the first time in the 80s [31]. A general overview of some polymeric hernia-repair meshes commercially available is reported in Fig. 2.