Tissue engineering is defined as an interdisciplinary field that applies the principles of engineering and the life sciences toward the development of biological substitutes that restore, maintain, or improve tissue function [1]. These substitutes contain a cell component and a material component, which serves as a carrier for the cells. In advanced tissue replacements, the material component should actively promote the adhesion, growth, differentiation, phenotypic maturation and other cell functions in a controllable manner. In other words, the material component should act as an analogue of the native extracellular matrix.

In the first part of our studies, we focused on endothelialization of clinically-used knitted polyethylene terephthalate prostheses. The inner surface of these prostheses is rough and highly hydrophobic, and this hampers the adhesion and growth of endothelial cells. However, when covered with fibrin-based films (Fig. 1A), this surface allows the formation of a confluent and shear stress-resistant endothelial cell layer with well-developed cell-matrix contacts, namely talin-containing focal adhesion plaques (Fig. 1B) and intercellular contacts, visualized by staining of VE-cadherin (Fig. 1C) [2, 3].

In the second part of our studies, we concentrated on bone implants. Metallic materials are still unavoidable in load-bearing applications, e.g. as substitutes for big joints (hip, knee, shoulder), although these materials are too stiff, too weighty, and often release cytotoxic and immunogenic ions. In order to increase their chemical stability and their attractiveness for cell colonization, the surface of these materials is modified by various approaches, such as electric discharge machining, acid etching, shot peening or deposition of various films (Fig. 2A). Simultaneously, three-dimensional degradable porous or fibrous synthetic polymeric scaffolds have been developed, enabling the ingrowth of cells and the formation of newly-regenerated bone tissue (Fig. 2B, C) [4, 5].

The third part of our studies focused on structures for skin reconstruction and regeneration, based on degradable polymeric nanofibrous meshes. These meshes enable the adhesion, growth, phenotypic maturation of keratinocytes (Fig. 3 A, B), and enable them to communicate with the underlying dermal fibroblasts (Fig. 3C) [6].


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Fig. 1: A nanofibrous fibrin layer for inner modification of vascular prostheses (A), and immunofluorescence of talin (B) and VE cadherin (C) in endothelial CPAE cells in cultures on these layers. Bar 10 μm (A) or 25 μm (B, C).

Fig. 2: Human osteoblast-like Saos-2 cells immunostained for talin (A) and MG-63 cells (B, C) in cultures on Ti-6Al-4V alloy modified by electrical discharge machining and shot peening (A), and on porous (B) or fibrous (C) polylactide-co-glycolide scaffolds. Bar 30 μm (A) 400 μm (B) and μm (C).

Fig. 3: Immunofluorescence of cytokeratin 5 (A) and filaggrin (B) in human HaCaT keratinocytes in cultures on polylactide nanofibrous membranes treated with plasma (power 75 W, exposure time 30 s). C: Layers of HaCaT keratinocytes (green) and human dermal fibroblasts (red) separated by a nanofibrous membrane. Bar 25 μm (A, B) or 100 μm (C).