

Tissue Engineering

Giacomo Fantoni

telegram: @GiacomoFantoni

Ilaria Cherchi

Telegram: @ilariacherchi

Elisa Pettinà

Telegram: @elispettina

Gaia Faggin

Telegram: @GaiaFaggin

Github: <https://github.com/StefanoCretti/TissueEngineering.git>

June 15, 2022

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Chapter 1

Basics of tissue engineering

1.1 Definition of tissue engineering

1.1.1 First definition

At first, **tissue engineering** was defined as "*A combination of principles and methods of life sciences with that of engineering, to develop materials and methods to repair damaged or diseased tissues, and to create entire tissue/organs replacements*" (1980s). This definition was soon outdated since it did not consider a fundamental difference:

- **Repair** is the act of closing the wound, restoring the macroscopic structure; the tissue formed during repair is often different from the starting tissue (scar tissue for instance), and thus the different properties undermine the function of the repaired area due to the different properties.
- **Regeneration** is also a way of closing a wound, but the structure is restored using cells of the same type as the starting one, therefore maintaining the functionality of the area.

Tissue engineering in fact aims at regenerating the structure and functionality of the district, not merely closing the wound.

1.1.2 Regeneration-centric definition

A following definition puts the focus on the regeneration aspect, underlining the necessity to understand the mechanism guiding regeneration: "*The applications of principles and methods of engineering and life sciences, to obtain a fundamental understanding of structural and functional relationships in novel and pathological mammalian tissues, and the development of biological substitutes to restore, maintain or improve tissue function*" (late 1980s).

1.1.3 Current definition

Nowadays, tissue engineering could be defined as "*a biomedical engineering discipline that uses a combination of cells, engineering, materials, methods and suitable biochemical and physicochemical factors to restore, maintain, improve or replace different types of biological tissues*." Tissue engineering holds promise of producing healthy organs for transplant by using patient cells (or immuno

1.2. HISTORY OF TISSUE ENGINEERING

compatible cells). Tissue engineering can be combined with gene therapy, therefore including the correction of incurable genetic defects (such as sickle cell anemia).

1.2 History of tissue engineering

Tissue engineering is born from the concept that interaction between cells and extracellular matrix is important for understanding the structural and functional relationship of these components. The first experiment on tissue engineering was conducted by W.T. Green, who tried to regenerate bone using chondrocytes (since in the physiological process, bone is obtained by calcification of cartilage); he managed to obtain bone formation in nude mice, thus he concluded that with the advent of innovative biocompatible materials it would be possible to generate new tissue by seeding viable cells onto appropriately configured scaffolds. Later on, Langer and Vacanti created artificial scaffolds for cell-delivery, rather than using natural derived scaffolds which are difficult to replicate. The use of artificial matrices specifically designed for the system allowed to obtain reproducibility and high-quality. Tissue engineering holds promise of producing healthy organs for transplant. Its main aim is to grow tissues or organs *in vitro* or designing matrices able to induce regeneration *in vivo* by recruiting patient cells.

1.3 The need for tissue engineering

Tissue/organ transplant is a heavily limited solution for tissue/organ failure; some of the main limitations being:

- Donor-recipient compatibility: it is almost impossible to find a fully compatible donor (all major histocompatibility complexes matching with the recipient) and the use of non-fully compatible organs/tissues requires the recipient to undergo immunosuppressive therapy (generally chronically) to avoid rejection of the transplant.
- Rejection risk: rejection can occur regardless of compatibility and immunosuppressive therapy, therefore this risk can never be avoided completely.
- Organ/tissue scarcity: even not considering compatibility, the amount of organs/tissues that can be donated is very scarce, since most of them come from car accident victims or relatives.

For this reasons implanting artificial devices has grown more popular since:

- They are ready to use.
- They immediately restore the function of the organ/tissue.
- They can be personalized.
- They do not cause rejection.

Still, implants present many limitation:

- They require invasive surgery.
- They can cause foreign body reaction.
- They cannot replace completely the functions of the organ/tissue due to limited performance.
- They have limited duration.

1.4. TISSUE ENGINEERING PARADIGMA

1.3.1 Examples of implant devices

Some examples of implant devices are:

- **Hip joint prosthesis:** made of metallic alloys (mostly based on titanium) and ceramic (for the joint socket). It immediately restores the function of the joint but it requires very invasive procedures (long segment inside the femur) and overtime it sticks to the bone, making it really hard to substitute it. This last point is of little relevance considering older people, but it is a big problem for younger patients.
- **Vascular stent:** a stent is a cylindrical tool that is used in angioplasty procedures, meaning it prevents the stenosis (blockage) of usually coronary arteries. A flexible probe mounted with a stent, a balloon and some way to visualize the probe from outside the body (via ecography for example) is inserted in the femoral artery; then the probe is navigated to the damaged region and the balloon inflates positioning the stent. This allows to restore blood flow to the myocardium without open heart surgery, even in an emergency setting such as in case of heart attack. The main problem is that the artery keeps contracting and expanding, therefore the stent can damage the vessel causing necrosis, proliferation of fibrotic tissue and inflammation; one way to reduce the problem is to use polymers rather than metallic alloys, since they are more flexible, but they are also less durable. Moreover stents are in contact with the blood and thus provide an abnormal surface that can start platelet aggregation and coagulation, creating blood clots; the use of slow release anticoagulant drugs stents helps with that aspect.
- **Artificial heart:** heart shaped device with valves. This device does not work as a pump since it cannot contract like myocardial fibres, therefore it requires some form of auxiliary external pump. Just like with stents it can cause abnormal coagulation on the surface of the device or turbolences in the blood flow.
- **Bone scaffolds:** they are three-dimensional biomaterial structures used for bone defect reconstruction. An ideal scaffold should have features such as improving cell adhesion, proliferation, osteogenic differentiation, vascularization, host integration and, where necessary, load bearing (drugs for instance). These design parameters should lead to specific scaffold properties, which include biocompatibility, porosity, micro and nano-scale structure, degradation rate, mechanical strength, and growth factor delivery, all of which dictate the biomaterial to be used or developed.

1.4 Tissue engineering paradigm

Since artificial devices do not replace all the functions of a lost organ or tissue and often fail in the long term, a new approach stems from two considerations:

- Living tissues and organs can be routinely assembled and reliably integrated to the body to restore, replace or enhance tissue and organ functions.
- Biomaterials can interact with living tissue

and influence cell function and response, moreover they can help when the patient lacks excess tissue necessary for transplants, overcoming the problems of autografts and allografts.

This approach is in fact tissue engineering which is, using yet another definition for it, "*creation of a new tissue for therapeutic reconstruction of the human body, by the deliberate and controlled*

1.4. TISSUE ENGINEERING PARADIGMA

stimulation of selected target cells, through a systematic combination of molecular and mechanical signals". Following this approach, instead of creating ex novo a tissue or organ to implant, some biological mechanism is induced to help the body heal itself. This idea is more generally represented by the term **regenerative medicine**, which includes not only tissue engineering but also cell therapy and gene therapy. The main actors involved in this process are:

- **Cells:** generally stem cells immunologically transparent derived from the patient to avoid rejection and able to differentiate into differing lineages for specific structural applications.
- **Scaffold:** the structure used to induce and guide the growth of the cells. Tissue engineering focuses mostly on material, surface and properties of this component. They need to be developed to be compatible with living systems and cells and their interface with the latter understood to optimize the regeneration process. These systems can be closed, semi-permeable or open and could incorporate the biologic signalling necessary for cell differentiation and function. The 3D structure of scaffold could be universal or custom-developed to one patient.
- **Time:** required in order to induce and obtain regeneration in unnatural conditions. Some tissues can be driven to competition in vitro in bioreactors, however, optimal incubation times vary from tissue to tissue. The new tissue will require an intact blood supply at the time of implantation. Moreover the state of development of the individual need to be understood, the age of the patient from which stem cells are isolated and how the biochemical characteristics of the implants will change over time.

1.4.1 Scaffold characteristics

The main characteristics of a scaffold that must be considered are:

- **Mechanical properties:** ability to withstand mechanical stress (elasticity, compressibility...).
- **Morphology:** shape, size, structure...
- **Physical properties:** behaviour when temperature, pH and other aspects of the environment change.
- **Histology:** type of tissue it has to replace.
- **Porosity:** permeability of the structure to different elements; it must allow the cells of interest to grow and penetrate into the structure while keeping outside unwanted cell types.
- **Water content:** the structure must contain enough water to allow nutrients to reach all the cells.
- **Surface:** the surface of the scaffold must be functionalized with molecules that are recognized by surface receptors of the target cells, thus inducing gene activation and some form of response (adhesion, expression upregulation/downregulation...).

1.4.2 Tissue engineering pipeline

From these premises we get the so called **tissue engineering paradigm**, which is the basic flowchart that most tissue engineering procedures follow. In general, the main steps are:

1.4. TISSUE ENGINEERING PARADIGMA

1. **Collecting and isolating host cells of interest:** the type of cells needed for the procedure depends on the damage site. According to the cells needed (generally stem cells, sometimes primary cells) a biopsy is performed on an adequate tissue. Moreover, since tissue are generally heterogeneous, some steps are required to isolate the cells of interest from the others. Since donor and recipient coincide, there are no compatibility issues.
2. **Seeding cells on a scaffold:** the cells are then seeded on a scaffold made of some biomaterial which is biocompatible for the application at hand. This scaffold must provide adequate conditions for cell growth and proliferation, such as the presence of growth factors and cytokines.
3. **Cell stimulation in a bioreactor:** the seeded scaffold is then placed into a static or dynamic bioreactor which induces and stimulate cell growth and proliferation. This bioreactor must be able to provide all the stimuli needed for cell differentiation and organization according to the desired final result (this may include mechanical stress for instance, needed for myocardial differentiation, or type of surface, since the differentiation of chondrocytes depends on the form they assume due to adhesion).
4. **Re-implantation:** the fully prepared and populated construct is then implanted into the damaged area, where it will integrate itself with the surrounding tissues.

This approach has some major downsides, namely:

- It is very **labour intensive** and **time consuming**, since it takes time for the cells to proliferate and populate the scaffold. Moreover, the growth conditions depend on the cell type of interest (which can be difficult to define, since the whole physiological environment must be taken into account, therefore angiogenesis, immune system, vacularization, lymphatic system and much more).
- Given the production time, this approach is **not ready to use**, therefore it cannot be used in an emergency setting.
- The amount of time and labour needed for the production implies **high cost**.

1.4.3 In situ regeneration

In some cases, it is possible to simplify the procedure by implanting the scaffold immediately after it was populated with cells in the damaged area; this is called **in-situ regeneration**. In situ regeneration requires way less preparation time and uses the body of the recipient as a bioreactor, therefore reducing the labour needed for setting up the bioreactor setup and the overall cost. Notice that animal models cannot be used as bioreactors, since the mechanical properties of the tissues are not comparable (vertebrae of pigs and sheeps are built to support different weights compared to the human ones), the use of primates is strictly regulated by law and no animal is perfectly compatible with humans introducing so a risk of rejection. The in-situ regeneration approach is not always applicable and just like the slower version it has a lot of room for improvement, for instance the development of more functional and durable polymers to use as scaffolds. Another problem for both strategies is the need of a starting material, generally stem cells: the patient may not have enough excess tissue for an autologous transplant for instance. One way to mitigate this problem would be to save part of the umbilical cord to extract staminal cells, but in Italy this is forbidden by law.

1.5 Tissue engineering approaches

There are two different **approaches** for tissue engineering:

- **Top-down approach:** this is the traditional approach: cells are harvested from the donor, cultured and modified if needed. They are then seeded on a porous scaffold that during cell proliferation is slowly degraded by the cells and replaced by extra cellular matrix (ECM). The engineered tissue is then implanted into the patient. The main advantage of this procedure is that it is possible to produce mechanical stress, which is needed for the differentiation on certain types of cells.
- **Bottom-up approach** (modular approach): some fundamental elements, such as cell sheets, cell aggregates, cell laden modules and bioink (3D printer ink containing cells) are used to construct a 3D module assembly, which can then be implanted into the patient. This modular building process allows to create very complex structures, with gradients and without a scaffold. Since a temporary gelatinous matrix is used, the module assembly lacks the rigidity provided by a scaffold, therefore it does not easily maintain the mechanical stress. Furthermore, when 3D printing, many other aspects have to be taken into account, like the permeability of the matrix to nutrients, the sensibility of the cells to the stress due to the extrusion from the needle and others.

Chapter 2

Biocompatibility

2.1 Introduction

Biocompatibility is an essential aspect to take into consideration for the specification of the medical device that is being designed. Before building a scaffold or a biological device its time, location and individual in which it will be used need to be defined. A scaffold to be useful should activate specific cellular function and reflect and exploit the different mechanical and chemical properties of the tissue in which it will be implanted. Because of this a scaffold it's designed taking into account the specific region in which it will be implanted considering the cell population and kind of injury among other parameters. Cells are the building blocks of biology and they can read the information coming from the external environment. Specific gene expression is activated in cells given external signalling molecules. This makes it evident that the environment reaches the cell through chemical and mechanical stimuli. Because of this scaffold are not self-sufficient entities: they are bioactive and need to collaborate with environment, cells and the extracellular matrix to perform their function.

2.1.1 Parameters defining biological outcome

The biological outcome that will be obtained then depends on a list of parameters which, once defined allow for the creation of a scaffold capable of regenerating the injured tissue. These parameters are:

Porosity is extremely important for cell migration. The material should be 3D for the cells, meaning that the scaffold should allow and promote cell adhesion, growth and migration.

Mechanical properties are different for each tissue, the scaffold should both resist physical stress and provide the correct stimulus that the cells need to grow, adhere, migrate and differentiate.

Surface modification is often referred to the functionalization of the scaffold through the addition of proteins or sequence of aminoacids to increase adherence.

Antibiotic/antiviral drugs release system, to control chronic inflammation and possible infections.

Surface topography may be smooth or rough for example.

2.1. INTRODUCTION

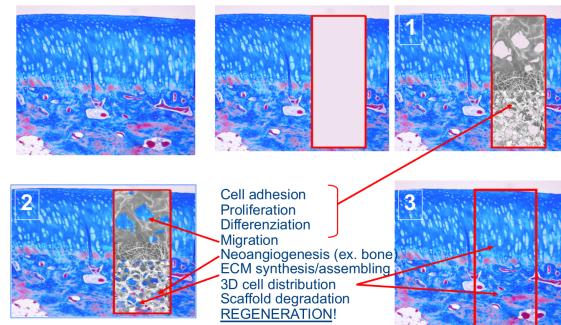


Figure 2.1: Cartilage regeneration

2.1.2 Guided tissue regeneration - cartilage example

In figure 2.1 we can see an articular cartilage, an intermediate tissue and bone. These three different tissues present differences in vascularization, cell organization (in cartilage cells live in lacuna, where migration and proliferation are downregulated) and innervation.

In case of trauma most of the time the damage reaches the bone to the lack of innervation in the cartilage. Because of this a multicomponent scaffold that accounts for different types of tissues is designed. In particular the scaffold should:

- Upregulate angiogenesis in the bone and downregulate it in the cartilage.
- Increase the water content in cartilage so to increase its compressive capabilities through hydrophilic materials.
- Provide an environment for osteoblasts and lacunae for cartilage cells
- Provide space for nerves, vascularization and cell migration.
- Tissue specific degradation times.

After the design biocompatibility needs to be checked: it will have an impact on cell adhesion, proliferation, differentiation and migration, on neoangiogenesis and on the regeneration of the function of the damaged tissue.

2.1.3 The principles of tissue engineering

A tissue engineering process is composed by three steps:

- Design.
- Material choice.
- In vitro and in vivo testing.

Tissue engineering implements a multidisciplinary strategy and the advance in materials' science and biology drastically improve the success of scaffold application. The advances in biology, for example the discovery in 2000 of macrophages $M1$ and $M2$, allow for the design of more biocompatible scaffolds.

2.1.3.1 4th generation biological regenerative biomaterials

A biomaterial to be defined as a 4th generation biological regenerative biomaterial needs to:

2.2. DEFINITION

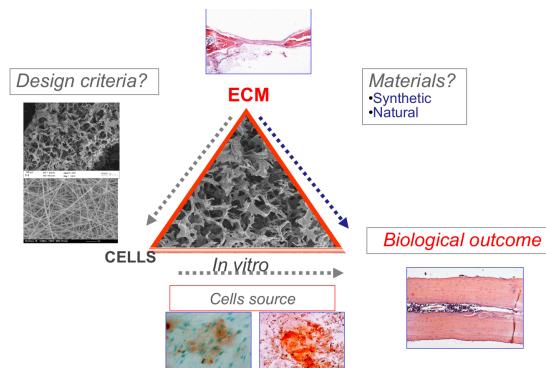


Figure 2.2: Tissue engineering principles

- Change its activity based on the environment: be, for example thermo or pH responsive.
- Be instructional: functionalized with peptides that can control cells' fate.
- Have specific mechanical strength and functions to allow for mechanical signalling.

Biomaterial need to be stable and inert in the beginning to allow for drug delivery, for printing organ and for cell therapies. The hope for the future is to develop also diagnostic systems and to implant electronic devices.

2.2 Definition

2.2.1 First definition

At first a material was defined as biocompatible when inert, presenting a total absence of interaction between the material and the tissues. Minimal reaction to the foreign body was preferred, meaning no (or low) inflammatory reaction and no immuno-response. This definition focuses on the body reaction to the implant, requiring on the latter only chemical stability.

2.2.1.1 Problem with the first definition - heart valves

The problem with this definition were brought to light when observing the implant some time after the implantation. We will now discuss as an example the problem with heart valves. An heart valve made of titanium and polyester failed because around the plastic layer the body built a very thick scar tissue, not letting it open anymore. The same valve with an ultrathin silver layer used for antibiotic purposes caused thrombosis in other patients. To tackle this problem this valve was treated with anticoagulants. A biological heart valve harvested from pigs failed because of calcification which rendered the valve unable to close.

2.2.2 Second definition

All the material induce a biological reaction. No material is completely inert, because our antibodies can recognize anything that is not "self". The concept of biocompatibility was revised as the ability of a material to perform a specific application causing an appropriate host response in a specific

2.2. DEFINITION

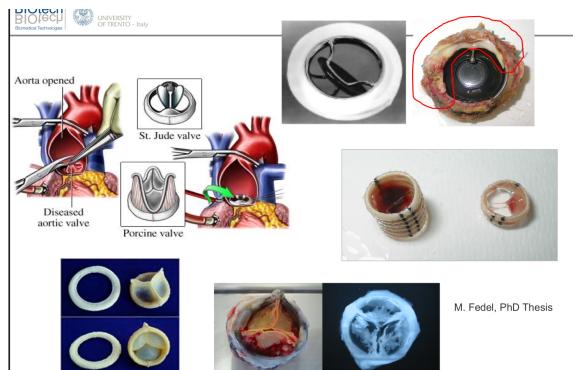


Figure 2.3

application (1987). It is clear how the context of implantation is fundamental: scaffolds should be tissue and organ dependant, resolving a defined situation by reacting with the tissue and activating the cellular function to heal a specific environment, before being degraded.

2.2.2.1 FBRx - an example

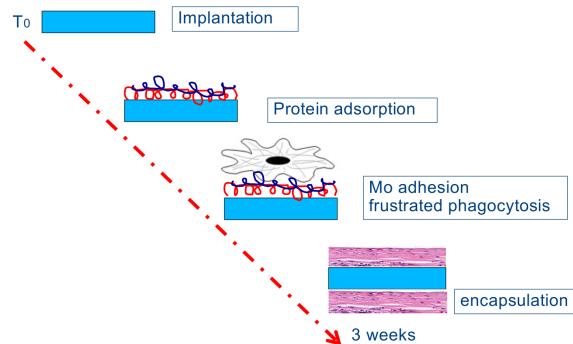


Figure 2.4

We can take as an example FBRx, a non porous scaffold material. Depending on the chemistry of the material activates a specific protein adsorption and subsequently a specific immune response. If the material is not degradable, the macrophages start to coat the foreign body with scar tissue, ending the adsorption reaction, allowing the scaffold to degrade, leaving an empty bag of scar tissue. Changing the porosity the scar tissue is reduced and regeneration is promoted. For this case injectable gels (hydrogel) are useful scaffold: they are porous and leave space for cell migration. Because of this they are used when filling cavities.

2.2.3 Re-evaluation of the biocompatibility concept

Some factors led to the redefinition of the concept of biocompatibility, as for the scaffold is not enough to simply exist.

2.2. DEFINITION

- Response to specific materials could vary from one application site to another, it is tissue-organ dependent: there is a need to define both material characteristics and implantation context, composed by the tissue and the injury.
- An increasing number of applications re-

quire that the material specifically react with the tissues rather than be ignored by them.

- Some applications require that the material degrade over time in the body rather than remaining indefinitely in it.

Because of this biocompatibility can be considered a system's property: it cannot be defined without considering the implantation context: it involves the separate, but interrelated, responses of the two phases of the biomaterial-tissue complex and interfacial phenomena triggered by their compact. The key to understand biocompatibility is:

- The determination of which chemical, biochemical, physiological, physical or other mechanism become operative and why.
- The determination of the highly specific

conditions associated with contact between biomaterials and tissue of the body.

- What are the consequences of these interactions.

2.2.3.1 Mechanotransduction

Mechanotransduction describes the processes at a cellular and molecular level that are involved with the transduction of mechanical stimuli into biochemical signals: structural and hemodynamic forces are encountered at the interface between tissue and scaffold. There will be a mismatch of elastic moduli between tissues and engineering material, causing differential stresses and strain. These processes will cause sensing and signalling processes to modulate gene and protein expression profiles.

2.2.3.2 Sterile inflammation

Sterile inflammation is an inflammation that results from trauma or chemically-induced injury without the involvement of any microorganism. It is associated with the recruitment of neutrophils and macrophages and the generation of pro-inflammatory chemokines and cytokines. The progress of biomaterial-induced sterile inflammation has to be considered during implantation. Among the processes involved into sterile inflammatory response there is:

- Pattern recognition receptors can sense conserved structural entities in microorganisms and exogenous molecules.
- Inflammasome activated by immune system molecules induce inflammation in response to pathogens and molecules derived from the host.
- Macrophages change phenotype according to the stress factors from anti to pro inflammatory situations depending on the damage associated molecular patterns.

2.2.4 The generic biocompatibility pathway

The simple generic pathway (figure 2.5) starts with the presentation of a clinical condition which leads to the decision to use a biomaterial-based therapy. The biomaterial may interact with cells in the determination of the appropriate host response. In most situations the desired clinical outcome can only be achieved through a combination of effects on critical cells and the avoidance of effects

2.3. THE COMPLEXITY OF THE BIOCOMPATIBLE SYSTEM

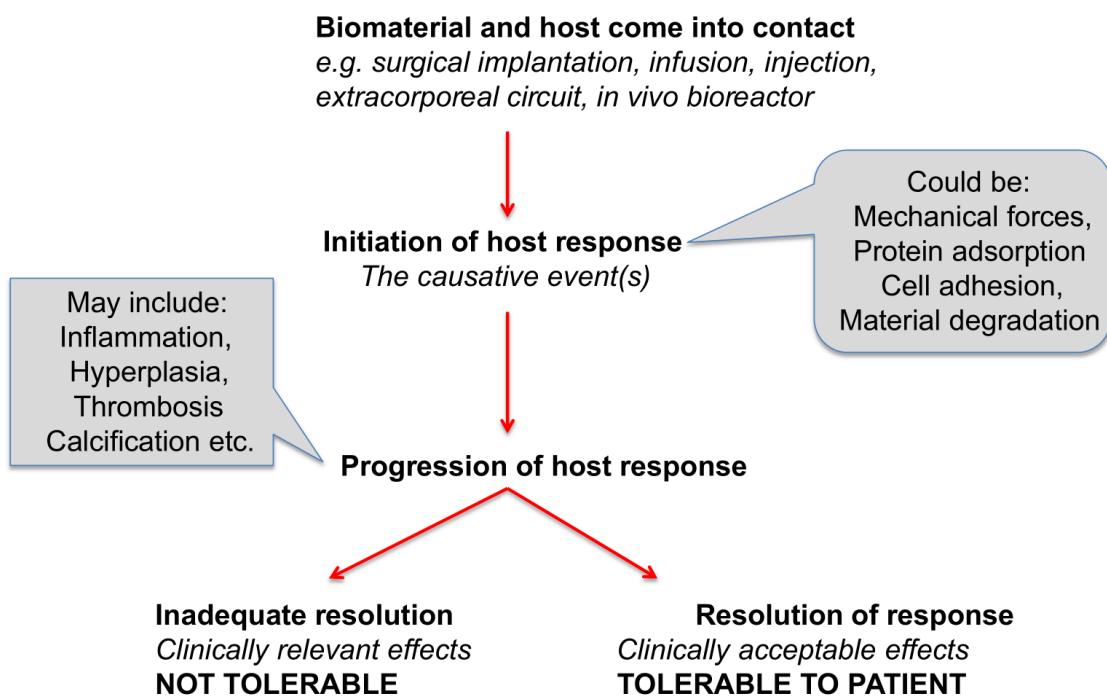


Figure 2.5: General biocompatibility pathway

on other cells. The part of the pathway between biomaterials and cells constitutes the generic biocompatibility pathway. The biomaterial will influence the events through mechanical or molecular signalling processes.

2.2.4.1 Cells

Cells on this framework (figure 2.6) can be:

- Target cells: cells at which the therapy is aimed.
- Defensive cells: cells of innate and adaptive immunity, whose job is to repel and remove injurious external agents.
- Interfering cells: cells that interfere with the response that the biomaterial is seeking.

The biocompatibility pathway will be determined by the events in these groups. The key to an appropriate host response is the dominance of the desirable effects over undesirable effects on the target cells, coupled with the avoidance of unacceptable responses via the defensive and interfering cells. The effects of chemical and mechanical events mediated by the biomaterial on cells are described on figure ??

2.3. THE COMPLEXITY OF THE BIOCOMPATIBLE SYSTEM

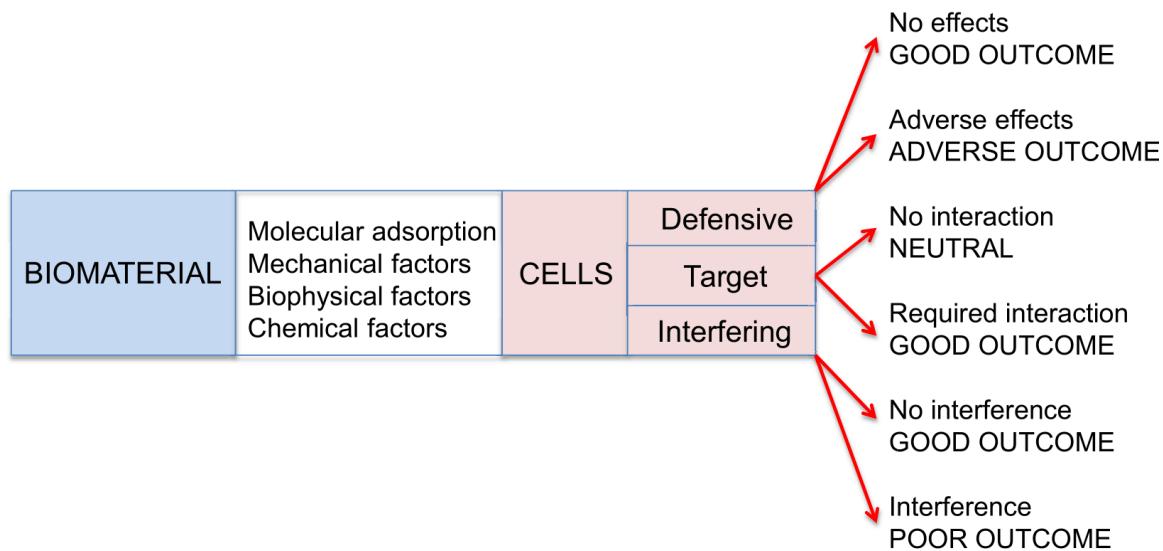


Figure 2.6: Cells involved in a general biocompatibility pathway

2.3 The complexity of the biocompatible system

As can be seen in figure 2.8 only three actors are present in the tissue engineering paradigm, but each one has many ramifications and possibilities.

To design a biocompatible scaffold there is a need to take into account:

- Determination of native tissue or organ context.
- In vivo testing with bioreactors that provide a dynamic environment and can provide mechanical stimuli by perfusion.
- Passage from in vitro to in vivo testing selecting an appropriate animal model.
- Definition of the surgical model and finally we move to clinical trials.

2.3.1 ECM molecules production: effect of the mechanical stimuli on a cartilage scaffold

Different sponge-like scaffolds were produced with different properties like porosity for cartilage regeneration. The quality of GAG (glycosaminoglycans) mainly into the ECM of cartilage was assessed. In addition microCT, a 3D imaging technique was used to see the level of infiltration of GAGs into the sponges. It can be seen in image 3.1 how SL500 had a lot of green, representing GAG, but only on the outside of the sponge. When the sponges undergo a mechanical stress similar to the biological situation the infiltration was present also in the inside of them. So, changing from a static to a perfusion environment the behaviour of the molecules and the sponges changed drastically. In particular during perfusion integrins were upregulated. These molecules are transmembrane proteins responsible for cell to cell and cell to extracellular matrix communication and imparted an instructive behaviour to the cells, which are extremely sensitive to it.

2.3. THE COMPLEXITY OF THE BIOCOMPATIBLE SYSTEM

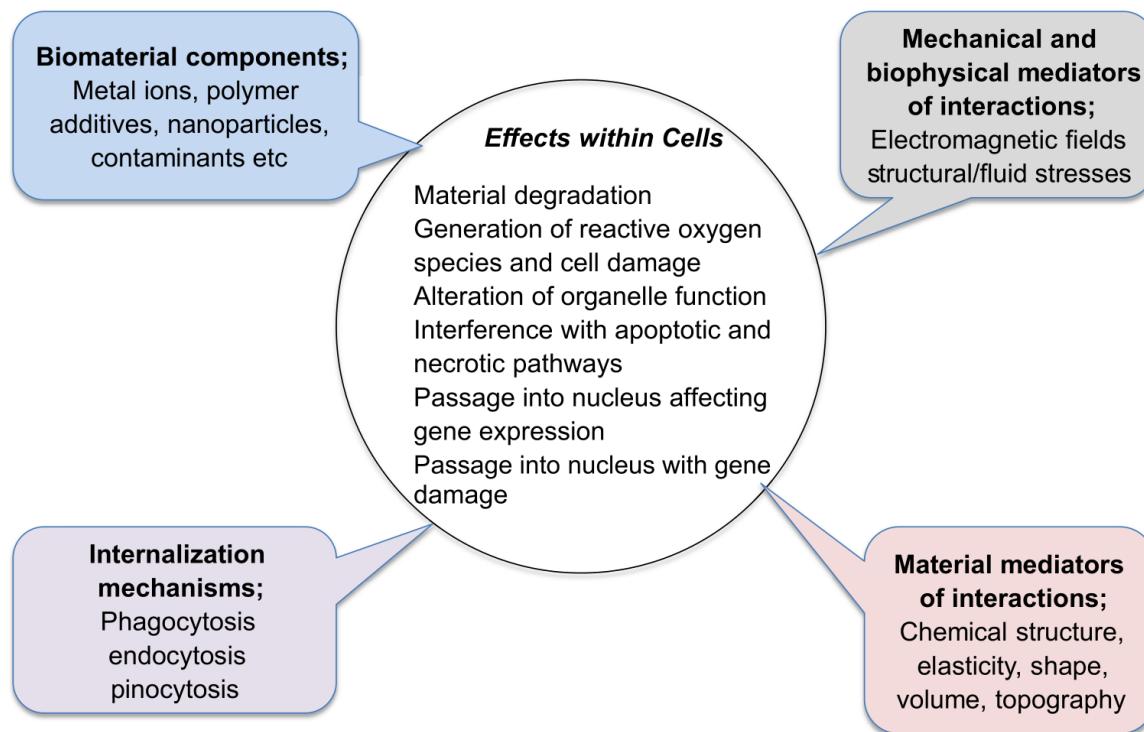


Figure 2.7: Effects of biomaterial-mediated events on cells

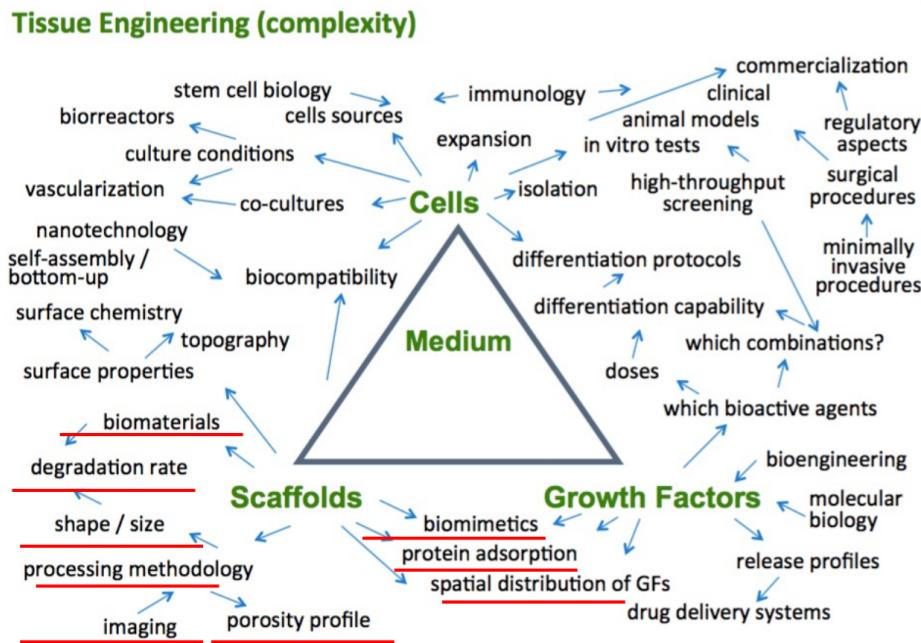


Figure 2.8: Tissue engineering can be represented considering three actors

2.3. THE COMPLEXITY OF THE BIOCOMPATIBLE SYSTEM

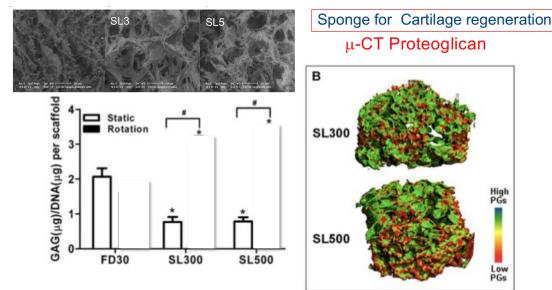


Figure 2.9: Cartilage sponges' quality assessment

Chapter 3

Extracellular matrix - ECM

3.1 Introduction

3.1.1 ECM functional role

The extracellular matrix provides a starting point for scaffold biodesign in tissue engineering. It is a dynamic environment produced and regulated by cells which performs different functions:

- Aids in cell locomotion through transmembrane receptors and ligands in the ECM. This adhesion ligands are patterned in order to direct cell locomotion in a specific direction.
- Transmits and distributes mechanical loads: the ECM assumes a specific structure in order to transmit mechanical stresses to the cell in a tissue-specific manner.
- Prevents premature mechanical failure.
- Partitions cells and tissues into functional units.
- Acts as a scaffold defining tissue and organ architecture.
- Acts as a storage and dissipative device for elastic energy.
- Acts as substrate for cell adhesion, growth, and differentiation. Adhesion is the activation step that allows for cell growth and differentiation, and because of this it fundamental for the activation of regenerating cells.

The ECM is a controlled complex network composed of proteins, glycoproteins and proteoglycans (PGs), which assemble in a tissue-specific manner to provide tissue specific biophysical and biochemical properties.

3.1.1.1 ECM-cell interaction

The interaction of the ECM with the cells is necessary as to avoid cells entering into a pathogenic state. Cell's nuclei and the ECM interact through integrins. This integrins convert external signals into a context-specific gene expression profile. The ability of cells to sense the chemical, mechanical and topographical features of the ECM enables them to integrate complex, multiparametric information into a coherent response to the surrounding microenvironment. Consequently, dysregulation or mutation of ECM components results in a broad range of pathological conditions. Because of

3.1. INTRODUCTION

this the ECM can be considered as a composite material where cells sense the environment. ECM polymers or networks are instructive: they provide structural and mechanical integrity to tissues. A tissue engineering scaffold should activate this integrins and ligands as to activate the gene expression profile necessary for regeneration. The focus should be on activating the correct pathway as to have the desired therapeutic effect and not a damage one.

3.1.1.2 Regulating cells' fate

The ECM controls whether a cell should undergo apoptosis or necrosis. Apoptosis is a programmed cell death, in which cells are encapsulated by vesicles and removed. Necrosis instead happens when the cellular membrane is damaged, causing the rupture of the cell and inflammation. During necrosis the activity of macrophages is increased, producing also inflammation for neighbouring cells. It has been suggested to induce tumour cell death through apoptosis to avoid upregulating inflammation.

3.1.2 Components of the ECM

The ECM is composed of different molecules:

- Physical signals: fibronectin, vitronectin, collagen, laminin, fibrillin, glycosaminoglycans (GAGs), PGs.
- Soluble signals: growth factors (GF), cytokines, chemokines, used for communicating with distant cells.
- Structure: fiber-based and function dependent composites.
- Water: tune mechanical properties to better support compressive stress. Highly present in articular cartilage, its content is controlled by GAGs and PGs.
- Mechano-transduction: translating mechanical stimuli.

3.1.2.1 Collagen fibers

The ECM is composed mostly of collagen fibres, which can be organized into:

- Fibrils, very elastic structures.
- Fibres: more thicker and stiff than fibrils.
- Bundles: composition of bundled fibres.

To better tune the mechanical properties without changing the ECM's chemistry the orientation of the fibres will change, depending on the external mechanical stimuli. Change the organization of the fibres cause modifications in water content and rigidity, causing a completely different signal to be sent to the cells communicating with the ECM.

3.1.2.2 Signals

Signals are fundamental for cell adhesion and must be considered when designing a scaffold together with ligands and collagen. To reduce scaffold complexity by decreasing the number of molecule necessary in a scaffold multifunctional polymers are being developed. The cross-talk between cells and the ECM allows for the control of many processes like:

3.2. THE MATRIXOME: CHEMISTRY OF THE ECM

- Pattern formation.
- Morphogenesis.
- Cell fate.
- Daily cellular processes.
- Wound healing.
- Tissue homeostasis.

3.1.2.3 Stem cells

Stem cells are uniquely capable of reproducing themselves through self-renewing and of differentiating into the different cell types necessary to perform a specialized function. They are fundamental in embryogenesis and a control between self-renewal and differentiation is fundamental in the healing process and homeostasis. Deregulating this process can cause tumorigenesis, degeneration or a pathogenic state. Because of this they are very sensible to the ECM and to the environment.

3.2 The matrixome: chemistry of the ECM

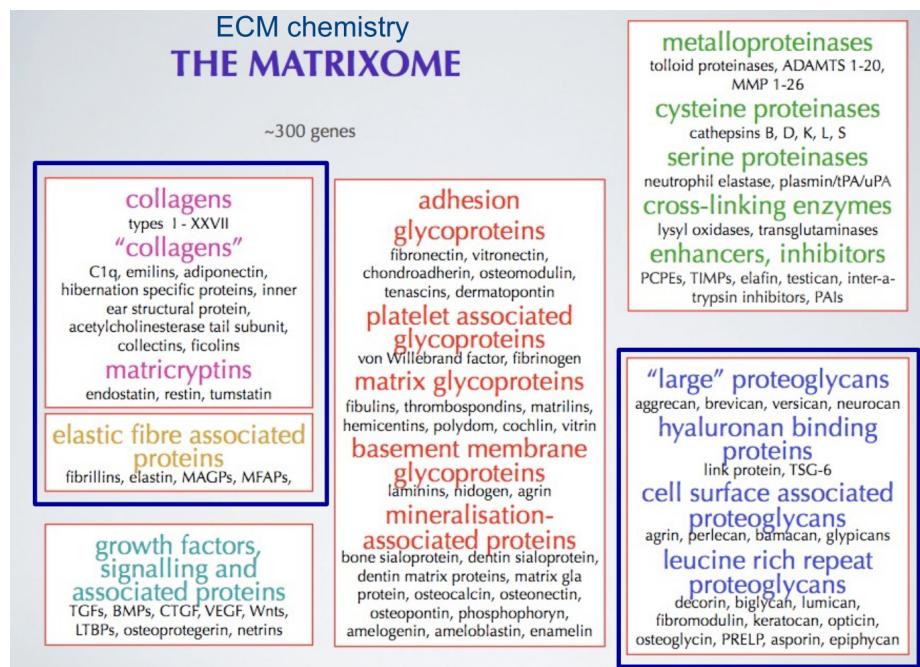


Figure 3.1: ECM content

The ECM is a mosaic structure that guides cell differentiation and function. It is a composite material that comprises a fibrous structural net embedded in a porous hydrated gel matrix of GAGs. Different properties of the ECM are controlled by different molecules, depicted in figure 3.1, present in it:

- Stiffness: collagens.
- Elasticity: elastic fibres.
- Cell-cell communications: signalling molecules like growth factors.
- Cell adhesion and mobility direction: adhesion molecules.

3.2. THE MATRIXOME: CHEMISTRY OF THE ECM

- Dynamic response: proteinases, which can be used to degrade the scaffold once it is no more needed.
- Water content: GAGs and PGs.

3.2.1 ECM - cell interaction

The cellular membrane is rich in proteins able to sense the external environment and transfer the information to the nucleus, activating a specific gene expression. When designing a scaffold, one must provide it the ability to interact with cells with an appropriate mechanism as not to deregulate cellular and tissue mechanisms.

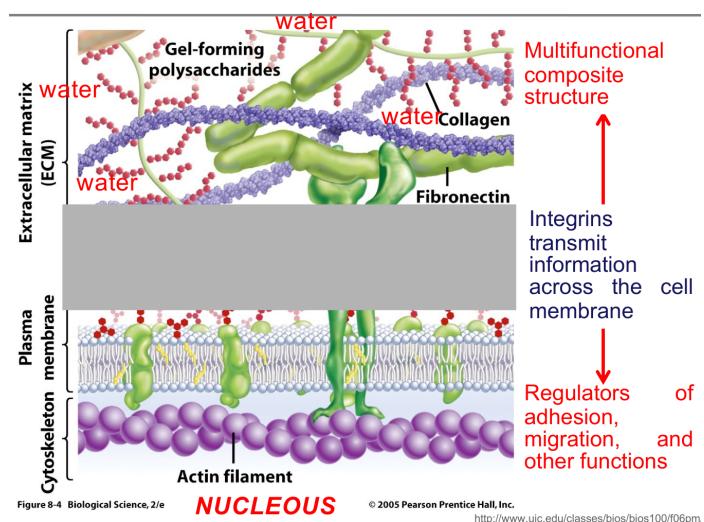


Figure 3.2

The ECM is a mosaic structure, a guide for cell functions. Natural ECM modulate tissue dynamics through their ability to locally bind, store and release soluble bioactive ECM effectors.

3.2.2 Matricellular proteins

Matricellular proteins are extracellular modulators of cell function expressed at high level during development and in response to injury. They are modulators of cell matrix interactions and bind to many cell surface receptors and to the ECM GFs. They are cytokines and proteases. Their role is to induce de-adhesion in contrast to the adhesivity of most matrix proteins allowing for cell migration. Other functions are:

- Cell adhesion, migration, chemotaxis.
- Matrix assembly and collagen fibrillogenesis, helped through some factors like ascorbate.
- Regulation proliferation/apoptosis.
- Binding/activation GFs and cytokines.
- Angiogenesis and tumour growth.

For example, TSP1 and 2, Tenascin-C and SPARC support a state of intermediate adhesion, with disruption of focal adhesion and the reorganization of actin stress fibres.

3.2. THE MATRIXOME: CHEMISTRY OF THE ECM

3.2.3 Collagen

Collagen is one of the most important proteins for the ECM, as it provides structural properties. In the ECM collagen assembles into fibrils, then fibres and bundles when needed. Collagen is a family of proteins with at least 23 different types. Depending on the tissue, there is a prevalence of one type on others. In the cartilage for example type II collagen is the most represented, while in tendons it is type III collagen, which is really similar to jellyfish collagen, currently commercialized.

3.2.3.1 Different collagen structures

The collagen can assemble in different structure, granting different mechanical properties to the ECM. For example:

- Connective tissue: less density, soft network, more water, fiber and fibrils.
- Meniscus: parallel fibers, more density, packed fibers, because of different functions, bundle (more stress).
- Myocardium: mechanical stresses, it should support the growth in one direction.

3.2.3.2 Cell-collagen interaction

Cells can migrate in a collagen low porosity material through a specific enzymatic process, which allows the opening of the collagen structure. A scaffold should be designed to promote cell migration.

3.2.3.3 Chemical characteristics and synthesis

Collagen is non-homogeneous, bottom-up, multi-functional, dynamic and has a hierarchical structure which is function dependent. It is composed by three chains that can form an helix formation. Controlling the chemical bridges between the chains the elasticity of collagen can be changed. Collagen production starts in the cell, where it is synthesized. Then selected prolines and lysines in it are hydrolysed. Later the hydrolysines are glycosylated, and three α -chain assemble, forming the procollagen triple-helix. This triple helix is secreted. In the ECM the propeptides are cleaved and the resulting collagen molecules can self-assemble into fibrils.

3.2.3.4 Dynamic nature of the ECM

The ECM is degraded by cell-secreted proteases and releasing bioactive components that remodel the collagen structure. Natural ECMs modulate tissue dynamics through their ability to locally bind, store and release soluble bioactive ECM effectors such as GFs.

3.2.4 Fibronectin

Fibronectin is a large matrix glycoprotein present in most body tissues' fluids. Functionally, it is the classic example of an adhesive glycoprotein, binding and interconnecting extracellular matrix components with each other and to the surface of the cells. It is one of the most important molecules through which cells interact with the surrounding environment. The binding of fibronectin to the cell surface's integrin receptors plays a critical role in cell migration during the development and postnatally. It also plays a role as chemo-attractant for cells in a damage site. Its fragments after a damage promote wound contraction.

3.2. THE MATRIXOME: CHEMISTRY OF THE ECM

3.2.4.1 Chemical characteristics

Fibronectin is a dimer composed of two identical chains connected by a disulphide bond. Different sections of fibronectin can be recognized:

- One can interact with fibrin and heparin.
- One can interact with gelatine and collagen.
- One can interact with the cell receptors through the arg-gly-asp ac. sequence.
- One can interact with the polysaccharide heparin.
- One can interact with fibrin.

3.2.4.2 Fibronectin interaction

Fibronectin is able to interact and crosslink different molecules as to transmit signals and stimuli in the ECM. This allows cells to respond quickly to changes in the environment. The RGD-loop in fibronectin is strategically placed to undergo strong conformational changes and constitutes a mechanosensitive switch for recognition by integrin receptors. Depending on the mechanical stimulus, fibronectin can assume specific conformations, changing its activity and, in doing so, changing the signal that it sends. In ECM we have fibronectins, collagen, gel-forming polysaccharides, water, actin and the cell (figure 3.2).

3.2.5 Proteoglycans and GAGs

Proteoglycans are composed of a head core protein and a chain of polysaccharides, which is hydrophilic (figure 3.3). The chain of polysaccharides is composed of glycosaminoglycans or GAGs, long linear polysaccharides consisting of repeating disaccharide units, composed of a uronic sugar and an amino sugar, which is usually sulfated. Proteoglycans are responsible for controlling the water content of the tissue. They aid the ECM in responding to mechanical stresses. Many proteoglycans can be linked together via long hyaluronic acid chains, forming giant complexes. In the cartilage matrix, for example, individual proteoglycans are linked to a non-sulfated GAG: hyaluronic acid to form a giant complex. Hyaluronic acid in particular creates cell-free space during embryogenesis filled with water into which cells can proliferate and migrate. Because of their negative charges, glycosaminoglycans and proteoglycans together control the water content of the tissue, determining the degree of "squishiness" of the matrix. They also allow for H_2O reentry after tissue compression. In fact, during compression, water gets squeezed out and the negative charges of PGs and GAGs draw Na ions, creating an osmotic pressure and forcing the escaped water back in when the compression force is no longer in place.

3.2.5.1 Core protein

Glycosaminoglycans are linked to a core proteins in proteoglycans. This core protein can be:

- Hyaluronic acid.
- Chondroitin sulfate.
- Keratan sulfate.

The core protein, which is hydrophobic is usually surrounded with hydrophilic molecules.

3.3. SOME EXAMPLES OF ECM FUNCTION DEPENDENT ORGANIZATION

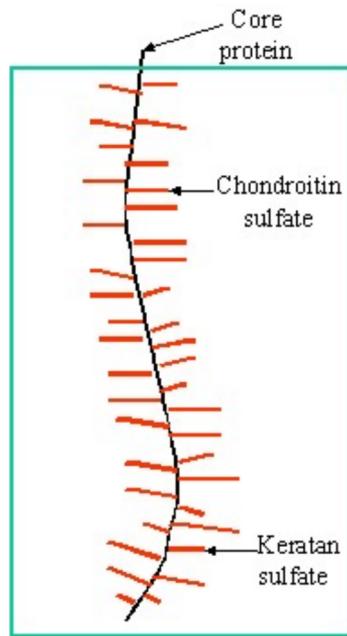


Figure 3.3

3.2.6 Elastin

ECM flexibility and extensibility are due to a network structure composed of elastin, a fibrillar protein that consists of 36 domains. This protein has alternating hydrophobic and crosslinks domain responsible for the long-range deformation of elastin and its high molecular mobility. It is composed during elastogenesis from the monomer tropoelastin released into the ECM. The interaction of elastin, that provides elasticity, and collagen, that provides stiffness, is fundamental for the mechanical properties of connective tissues. It can also transmit signals to the cells.

3.3 Some examples of ECM function dependent organization

3.3.1 Aorta wall

Considering the three main layers of aorta wall, the orientation and organization of collagen and elastin change, as structural adaptive response to mechanical requirements.

3.3.1.1 Mechanical requirements

Specific compliance and strength in both longitudinal and circumferential directions.

3.3.1.2 Structural strategy

Anisotropic and elastic multi-layers laminate, different fibre orientation through the thickness of the aorta wall.

3.3. SOME EXAMPLES OF ECM FUNCTION DEPENDENT ORGANIZATION

3.3.1.3 ECM main components

A 3D fibrous network of collagen combined with elastic fibres.

3.3.1.4 Sub-endothelium

Multilayered fabric of collagen. In a layer collagen fibres are uniformly oriented, while in different layers they have a different orientation to ensure proper resistance. Elastin is arranged following a 3D network of elastic fibres.

3.3.1.5 Media

The media consist of a complex network of smooth muscle cells, elastin and bundles of collagen fibrils. The fenestrated elastic laminae separate the media into several concentrically fibre-reinforced medial layers. This form alternating layers called musculoelastic fascicles. It is separated from the intima and the adventitia by internal and external elastic lamina, with a circumferentially oriented continuous elastin fibrous helix. The media can resist high loads in the circumferential direction.

3.3.1.6 Adventia

Collagen fibres are organized in thick bundles.

3.3.1.7 Intima

One layer of endothelial cells in contact with the blood flow, supported by an elastic lamina.

3.3.2 Bone

3.3.2.1 Mechanical requirements

High strength and rigidity combined with good toughness.

3.3.2.2 Structural strategy

In cortical bone, collagen fibrils are densely packed and form concentric lamellae. Fibrils in adjacent lamellae are arranged in perpendicular planes. Trabecular bone has a loosely organized porous matrix.

3.3.2.3 Composition and structure

The bone main function is to sustain and protect organ in the body and act as a storage for mineral homeostasis. It is characterized by a unique composition which gives rigidity and strength while maintaining elasticity. Minerals and collagens provide good hardness, compressive, tensile and torsion strengths. It contains osteoprogenitor, osteoblasts, osteocytes and osteoclasts. In cortical or compact bone collagen fibrils form concentric lamellae perpendicular in adjacent one. Trabecular or cancellous or spongy bone has loosely organized spongy arrangement.

3.3.3 Hyaline cartilage

3.3.3.1 Mechanical requirements

Flexible tissue, but strong and resistant to high pressure.

3.4. MODELLING NATURE

3.3.3.2 Structural strategy

Biphasic poroelastic medium with a heterogeneous structure: many layer with different properties and orientation of fibers.

3.3.3.3 Gradient structure

Collagen fibrils in the superficial zone are parallel to the surface to provide great tensile and shear strength. There are three zones defined by collagen fibres orientation: in the superficial they are parallel, in the transitional random and in the radial perpendicular. 80% of the tissue is water.

3.4 Modelling nature

When designing a scaffold the goal is to provide components that will drive tissue regeneration. The environment in which the scaffold will be implanted is dynamic, with water allowing for most of the interactions. These interactions regulate a lot of processes, in which there is cell fate, tissue formation and tissue regeneration. The scaffold needs to allow the cells to work in a natural way, and because of this needs to be modelled based on what is observed in nature. Living organisms naturally provide a multiplicity of materials, architectures, systems and functions, all resulting from a stringent selection process. So, a strategy to design a scaffold should include:

- Polymers able to integrate.
- Molecular synthesis, done by the cells, at very high level of organisation, allowing molecular recognition, multifunctionality, self-diagnosis and a destruction-recycling process.
- Structure dynamics through responsive polymers, allowing for self-assembling, molecule interactions, adaptation and self-healing.

The bottom-up approach is more bio-mimetic, as it reflects how nature works: self-assembling blocks, which assemble thanks to the environment to perform a function. The scaffold design should account for building a context-specific microenvironment.

Chapter 4

Damage response

4.1 Introduction

Biocompatibility is the ability of a material to perform with an appropriate host response in a specific application. The first system interacting with the scaffold is the immune system, which first gives an immuno response and then starts the healing process. The immune system is tissue specific and changes based on the pathogenic status of it. The loss of regenerative capacity of an individual throughout its lifespan is linked to the evolution of immune competence. Taking for example a wound of the skin, in a fetus complete regeneration without scar tissue formation can be observed, whereas in adults fibrotic healing takes place. Some parameters from the fetal models can be taken into account when designing a scaffold that will reduce the formation of scar tissue in adult patients. The relationship between tissue healing and the immune response is very complex: factors can have both a positive or negative effect, depending on the tissue and the organ involved in the injury and the life stage (that can be embryonic, neonatal or adult) of the patient. In order to prioritize healing the inflammation must be reduced as soon as possible.

4.2 Tissue damage

A damage to the tissue will cause bleeding, the accumulation of pathogens and cell debris at the site of the open wound. After a tissue suffers a damage the immune system gets activated and recruited to the damage location, causing an inflammatory response. This inflammatory response is a defence step and it can be splitted in sub-steps:

1. Platelet activation.
2. Coagulation cascade.
3. Inflammation.

After these steps the system will be activated for regeneration.

4.2.1 Timeline of tissue regeneration

In 3-7 days cell begins to migrate and proliferate in the wound site, promoting ECM and other molecule's synthesis and angiogenesis. External pathogens are dealt by scar tissue formation, blocking the entrance of the wound, while internal pathogen are dealt by inflammation increasing the

4.2. TISSUE DAMAGE

blood flow into the damaged site through angiogenesis. This overall process is controlled by the inflammatory system cells like macrophages coordinated by interleukines (IL-1 and IL-2) During this time period (processes visualized in 4.1):

- Free particles like cell debris or other external factors are opsonized with C3b antibodies for phagocytosis. This is done so macrophages can recognize pathogens through the attached integrins.
- The inflammatory reaction is elicited through factors acting on leukocytes, mast cells and endothelium.
- At the end complement-mediated cytology happens, destroying the pathogens.

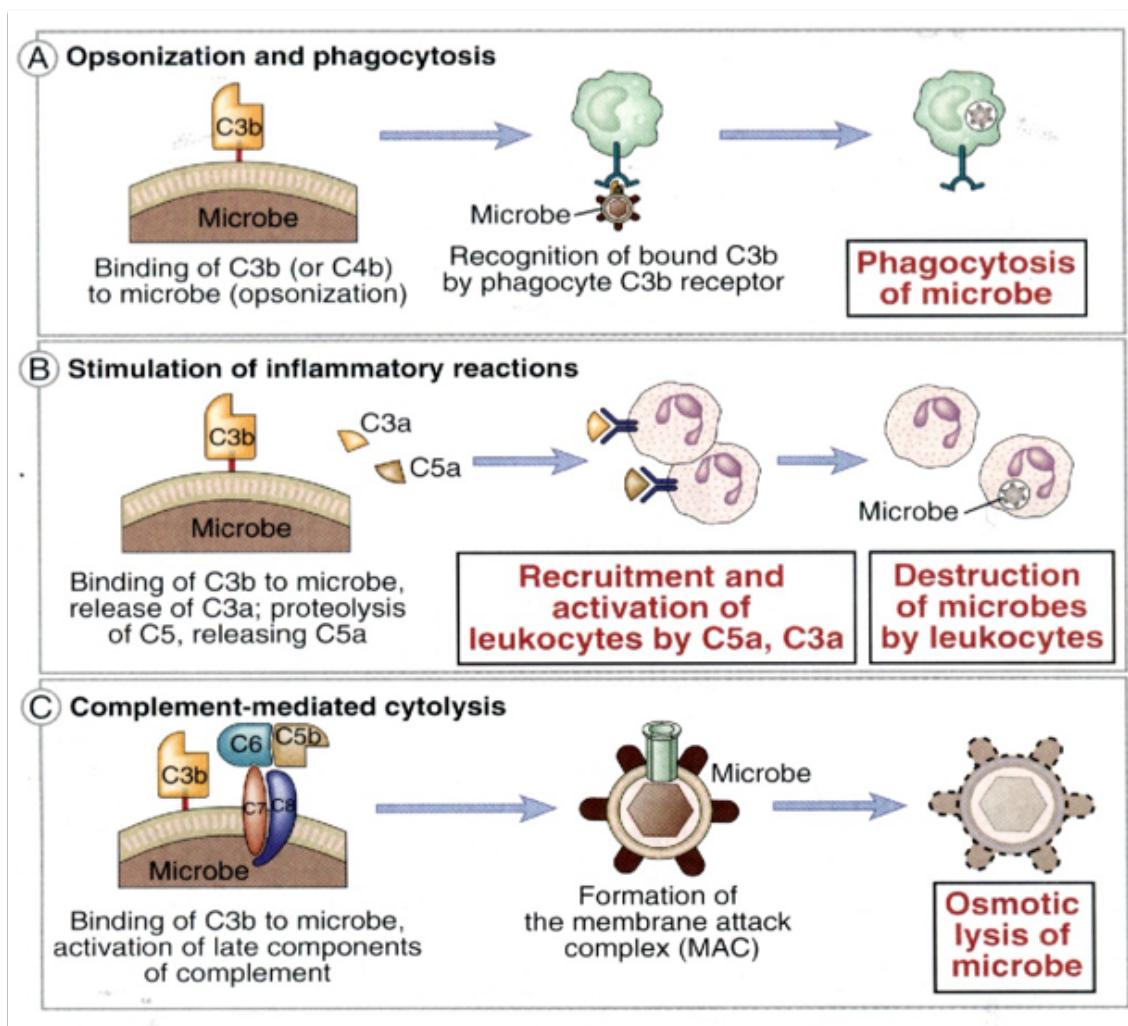


Figure 4.1

4.2. TISSUE DAMAGE

4.2.2 Immune system response

4.2.2.1 Inflammatory response

The inflammatory response is the body's natural response that occurs immediately following tissue damage. Its main functions are to defend the body against harmful substances and to promote the renewal of normal tissue. Signs of inflammation include:

1. Pain due to chemical released by damaged cells.
2. Swelling or edema due to an influx of fluid into the damaged region.
3. Redness due to vasodilatation (widening of blood vessels and bleeding in joint or struc-
- ture).
4. Heat due to an increase in blood flow to the area.
5. Loss of function due to increased swelling and pain.

The inflammatory reaction is the combination of a number of overlapping reactions within the body. Although a lot of these occur simultaneously, a certain order of events may be seen. The tissue damage may occur from trauma such as a tackle, collision or from a fall. However, quite commonly tissue injury is as a result of overuse (microtrauma) or pathology. So, when tissue cells become injured, a number of chemical like kinins, prostaglandin and histamine are released to initiate the inflammatory response. These chemicals work collectively to increase vasodilation (widening of blood capillaries) and the permeability of capillaries. This leads to increased blood flow to the injured site. These substances also act as chemical messengers that attract some of the body's natural defence cells through chemotaxis. Although highly beneficial to the body's defence strategies, some chemicals also increase the sensitivity of the pain fibres in the area and so the area becomes painful.

4.2.2.2 Immune system population

The immune system involved in response to a wound is composed by leukocytes which can be further divided into neutrophils and monocytes.

4.2.2.2.1 Neutrophils Neutrophils or polymorphonuclear are the most abundant circulating leukocytes. They are two types of intracellular granules, which are common progenitors for monocytes. They are produced on the number so 10^{11} per day, with a 6 hours lifespan, before apoptosis.

4.2.2.2.2 Monocytes Monocytes and activated macrophages are phylogenetically the oldest cells in the immune system. They circulate as inactive monocytes, and become activated macrophages when they enter a damaged tissue. They are typically located in:

- The subepithelial connective tissue.
- The interstitia of parenchymal organs.
- The lining of vascular sinusoids in liver and spleen.
- The lymphatic sinuses of lymph nodes.

They usually respond later than neutrophils at sites of injury and infection and they are the primary effectors of the innate immune system. The primary purpose of these phagocytes is to clear the wound from invading organism, everything that is non-self material. They can exist for years.

4.2. TISSUE DAMAGE

4.2.2.3 Phagocyte-mediated wound cleaning

Phagocyte-mediated wound cleaning is used to produce new vessels from new cells and it consists of different passages:

1. Recruitment: adhesion proteins facilitate attachment to endothelium.
2. Migration: receptors that mediate chemotaxis to target site.
3. Recognition and phagocytosis: specific receptors for microbes and opsonized materials driving phagocytosis. Fc receptors and C3 receptors are major mediators of the attachment.
4. Release of cytotoxic compounds: reactive oxygen and nitrogen species.
5. Cytokine production and secretion of numerous cytokines and chemokines with local and systemic activity:
 - Positive factors: increase macrophage activation, recruitment and stimulation of the adaptive immune system.
 - Negative factors: inhibit activation and proliferation of the immune system.
 - Secretion of factors that facilitate wound remodelling, matrix production and angiogenesis.

Macrophages dominate biomaterial interfaces in tissues and are often present chronically. There is a need to control the interaction of the scaffold with macrophages to avoid pro inflammatory signals.

4.2.2.4 Leukocyte extravasation

Leukocytes in physiological condition travel through the blood capillaries. When a tissue is damaged a chemotactic signal is sent to them through molecules like bacterial peptides causing their adherence to endothelial cells. Then integrins and I-CAM binds them causing adhesion and process formation, with additional signals and ligands promoting migration. The overall process is described in figure 4.2.

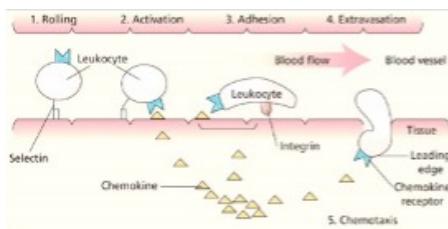


Figure 4.2

4.2.2.5 TNF- α

TNF- α is the principal mediator of acute inflammation. It is produced by macrophages and activates the pro-inflammation NFkB transcription factor. Doing so increases the endothelial expression of selectins and integrins, production of chemokines in endothelial cells and macrophages and the production of prostaglandins and leukotrienes like pyrogens and chemotactic. It also induced

4.2. TISSUE DAMAGE

prostacyclin expression in the endothelium, increasing local blood flow and production of IL-1 in macrophages.

4.2.2.6 Leucokyes migration

All of this signals cause chemotaxis of macrophages and neutrophils to the damaged area. Neutrophils are the first to reach the injured site and neutralize harmful bacteria. Macrophages aid the healing process by engulfing bacteria and dead cells and ingesting them, clearing the area and creating a favourable environment for new cells to grow. They can be found at the injured site within the first 72 hours and remain in it for weeks.

4.2.2.7 Major activities of macrophages secreted factors

The inflammatory system, which main goal is healing, pass through scar tissue formation, its modification and then regeneration. Macrophages react to inflammation by killing microbes and phagocytizing them, guided by oxotin produced by the system. The passage through scar tissue is necessary to defend the system from pathogens. So regeneration will happen only at a later stage.

4.2.2.8 Inflammatory monocytes

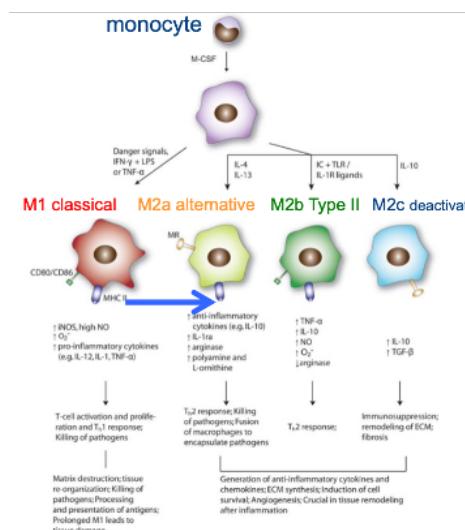


Figure 4.3: Monocytes differentiation process

Figure 4.3 depicts macrophages differentiation from classical proinflammatory to regenerative. Each one is characterized by a specific profile of released molecule, which can be used to identify them. This identification is useful to see the interaction between the scaffold and the inflammatory cells and see if they are able to produce angiogenic and regenerative factors.

4.2.2.9 Inflammatory cellular response

Opsonin allows the macrophages to recognize and digest a foreign body like cancer cells. This process involves moving it inside the cellular membrane of the macrophages. When a scaffold is introduced

4.2. TISSUE DAMAGE

into the injury site cells respond with:

- Oxidative burst.
- Bacterial killing.
- Tissue injury.

If the material is very sensitive it will release some fragments that will be destroyed by macrophages and will cause necrosis in the surrounding tissue. Doing so this particles will upregulate the inflammatory response, causing its transformation from acute to chronic. When designing a scaffold the aim is to reduce as much as possible the intensity and time of this response and, in doing so, the formation of scar tissue.



Figure 4.4

Figure 4.4 shows the interaction between cells and material in an implant, explanted after 12 years due to failure. The tube changed its shape and diameter, provoking blood turbulence, and increasing thrombosis risk. The problem with this implant is that space was occupied by macrophages, which were able to adhere and digest the external fibers causing their release in the form of microparticles. The prosthesis was made in nylon, a really stable polymer. Since it is stable and strong, the human body sensed it as foreign and started damaging it. The section of the damaged fibers was triangular. This continuous digestion process caused chronic inflammation, and, in the end, failure of the implant.

4.2.3 Wound healing in skin

After damage of the skin the regeneration process begins. This consists of:

1. Acute inflammation.
2. Proliferation.
3. Remodelling.
4. Restoring of the tensile strength.

The synthesis of collagen occurs between 7 and 14 days. After 14 to 21 days, collagen cross-linking assembling begins. During this process collagen has less cross-links, so it is more flexible but less strong. The immune system must be controlled as to induce tissue regeneration.

4.3. TISSUE HEALING

4.2.3.1 Macrophage transition

Right after the injury pro-inflammatory macrophages, phagocytes of type *I* will produce cytokines. When they remove all the debris type *I* macrophages differentiate into type *II* macrophages, leading to the downregulation of pro-inflammatory cytokines and tissue regeneration. When designing a scaffold this transition from type *I* to type *II* has to be carefully planned to shift from an acute inflammation to regeneration.

4.2.3.2 Main actors of the immune response

The main actors of the immune response following tissue injury are:

- Kinetic of immune cell mobilisation after tissue injury.
- Initial inflammatory phase following tissue injury.
- Immune mechanisms that can impair tissue healing or drive to scarring and fibrosis.
- Pro regenerative immune mechanisms.

4.3 Tissue healing

The process of tissue healing can be divided into four part:

1. Collagenation and cartilarisation.
2. Angiogenesis.
3. Proliferation.
4. Remodelling.

4.3.1 Revascularization

Revascularization happens when the damaged area begins to sprout new capillaries to bring blood to the regions once sufficient cleaning has been achieved. The new capillaries must sprout in an ordered manner, which must be taken into account during scaffold design. When blood flow has been re-introduced to the area, specific tissue cells begin to re-grow. For example, in a muscle tear muscle cells will repopulate the area. Wound healing occurs towards the end of the inflammatory process, however the two processes overlap considerably.

4.3.2 Collagenation

Macrophages work to clear the damaged area and make space for the regeneration of the new tissue. After a number of days, fibroblasts (collagen producing cells) begin to construct a new collagen matrix which will act as the framework for new tissue cells. Tissue healing occurs with the sprout of new capillaries to bring blood to the region (revascularization). A path for interconnection must be built by playing with scaffold architecture and the release of angiogenetic factors.

4.3.3 Prolifeareation

The proliferation phase lasts 4 weeks. In cases where the injury has been more severe, the affected area may be composed by a mixture between specific tissue cells (such as muscle cells) and other tissue known as granulation tissue. If this granulation tissue is not removed it will remain and form scar tissue, which can lead to a decreased functional ability of the tissue.

4.3. TISSUE HEALING

4.3.4 Remodelling

Remodelling occurs when new cells mould into their surrounding to once again produce a functional tissue. This process can take months or even years, altering the new tissue slowly. The new cells and protein fibers become arranged in a way that is best suited to the stresses imposed on the tissue. Hence, when a tissue is healing, it is important to stretch it in the correct direction so to optimise the strength of the new tissue.

4.3.5 Strategies to promote tissue regeneration

There are different strategies based on biomaterials and drug delivery systems to promote tissue regeneration by controlling the immune system:

- Physicochemical properties of the scaffold.
Degradability, hydrophobicity and topography must be kept in mind.
- Pro-inflammatory modulators.
- Anti-inflammatory modulators.

Materials must be chosen according to:

- Protein adsorption.
- Generalised toxic effect.
- Inflammatory cell activation.
- Fibrosis.
- Microvascular changes.
- Tissue-organ specific cell response.

The scaffold must activate or inhibit specific pathway, depending on the injury and on the tissue. These can be:

- Activation of clotting cascade.
- Immune cells response.
- Platelets adhesion, activation and aggregation.
- Hypersensitivity.
- Complement activation.
- Mutagenesis, genotoxicity.
- Antibody production.
- Tumour formation.

Everything is controlled by the material, the environment and their interaction. When the biomaterial is recognized as a foreign body, macrophages are recruited and cells will not be able to migrate on the scaffold for regeneration, resulting in a complete failure.

4.3.5.1 Effects of physicochemical modification to biomimetic scaffolds in musculoskeletal applications

In figure 4.6 the percentage of dendritic cells (inflammatory cells) in the sample can be identified. A different adhesion degree of inflammatory cells to the scaffold can be obtained changing the physical characteristics of the scaffold or the molecules with which it is functionalized. In panel B the different shape and thickness for each material can be seen. Moreover the control of dendritic cells' phenotype from pro-inflammatory to tolerogenic is of interest during scaffold design. Their phenotype is modulated by the innate immune cell types.

4.3. TISSUE HEALING

4.3.5.2 Processes that affect the biological outcome

The capability of a scaffold to absorb protein is important as it can cause an inflammatory response. In fact different protein conformation can control the water content, reducing it by increasing crystallinity, leading to different mechanical properties. It can be noted how a small amount of proteins is still able to induce a high pro-inflammatory reaction.

4.3.5.3 Empowerment of stem cells by inflammation

Immune cells at the site of tissue injury, including macrophages and T cells, secrete TNF α , TNF γ , IL-1, IL-13 and other pro-inflammatory cytokines, which in turn can activate stem cells. Once licensed by these cytokines, stem cells can facilitate tissue regeneration through cell differentiation and the release of anti-inflammatory cytokines and growth factors.

4.3.5.4 Designing immuno informed biomaterials matrices

When designing a functionalized scaffold a level of safety should be provided, non including tumorigenic factors. The molecules need to functionalize the scaffold can be delivered through extracellular vesicles. The topography and the chemistry of the scaffold play a central role in this process as macrophage polarization and network formation. The network should be formed in 3D on the surface, providing interconnections for macrophages. For instance, after scaffold implantation protein adsorption is observed, followed by macrophages adherence that cause the formation a network speeding up the process toward regeneration. Macrophages are also involved in vascularization. When the scaffold is not porous the macrophages will not be able to polarize and switch to state *II*, inducing the formation of scar tissue.

4.3.5.5 Effect of topography and stiffness on macrophages polarisation

Macrophages assume an elongated shape when switching to the M2 like phenotype with an up-regulated expression of arginase, on PDMS substrates with 2 gratings topography compared to planar control. After the adhesion (f), cells start to polarize (g) [figure 4.7]. Increased stiffness in hydrogel drives the transition in absence of cytokines stimulation.

4.3.5.6 Role of pore size, distribution and degradation

Figure 4.8 depicts silk fibroin micro-nets formic acid treated in an in vivo evaluation. In this case we have induced regeneration in unnatural conditions. Changing the pore size, the mechanical properties will be affected. In high porosity samples after 3 days we have granulated tissue (dots are cells), the scaffold was able to promote movement inside. At 10 days the arrows are indicating new capillaries. Macrophages are at phase II, cells have more space to build tissue. Then thinner fibers, blood vessels, not so many dots, signalling that inflammatory cells almost disappeared. Already after 3 days we have a very clear scar tissue formation, which becomes thinner in 10 days. We have also different vascularisation and degradation depending on porosity.

4.3.5.7 Regeneration of non-vascularized tissues

Since the mechanism of inflammation depends on the blood, non-vascularized tissues are generally hard to regenerate. When designing a scaffold different approaches are needed in these cases. An example is a damage to the cartilage-bone interface. The first is a soft, unnerve and non vascularized tissue, while the second one is a robust, innervated and vascularized one. In this situation we need a

4.3. TISSUE HEALING

gradient-rich scaffold that allows for correct tissue and interface regeneration, while avoiding blood vessel formation in cartilage that would drive the inflammation in the latter and transform it into bone. We can trap vessels in the bone section by playing with porosity, and we could tackle the cartilage section regeneration using injectable biocompatible gels filled with the appropriate cells.

4.3.6 Possible outcomes after the inflammatory response

After the intervention of macrophages and neutrophils and the death of neutrophils the immune system could face three different situations.

4.3.6.1 Clear situation

If the situation is clear: there is no more damage or pathogen that can be detected by immune cells and the remaining damage is of moderate extension. Local adaptive and innate immune cells produce anti-inflammatory cytokines IL-4, IL-10, IL-13 and TGF β leading to the definition of a resolving environment that leads to macrophage polarization towards the M2 phenotype. This macrophages appear elongated and rich in protrusions. This phenotypes communicate with local stem cells, mobilizing them and causing their proliferation and differentiation, and with local fibroblasts, triggering scar-tissue remodelling. In this process the temporary matrix is digested and a tissue specific ECM is deposited leading to healing by regeneration. This remodelling process can take months or even years.

4.3.6.2 Clear situation with extensive damage

If the situation is clear but the damage is too extensive, while the removal of the harmful stimulus leads to the release of anti-inflammatory cytokines and to the polarization of M1 (inflammatory and microbicidal) macrophages into M2 (immunomodulatory and reparative), and while these are successful into driving the local fibroblasts into a tissue remodelling status, the abundance of the scar tissue itself is a limit to the enzymes' proficiency: the scar tissue will remain, impacting the mechanical and functional characteristics of the tissue.

4.3.6.3 Not clear situation

If the situation cannot be cleared, after an appropriate period of time, usually one or two weeks, lymphocytes and plasma cells join M1 macrophages in a context characterized by cyclical instances of tissue healing by repair and tissue damage by immune cells' oxidizing activity. M1 macrophages are the vast majority, since the situation does not allow for the production of the anti-inflammatory cytokines needed for M2 polarization. There is the fusion of macrophages to form giant cells which recruits fibroblasts causing excessive collagen deposition and formation of fibrous scaffolds and scar tissue.

4.3. TISSUE HEALING

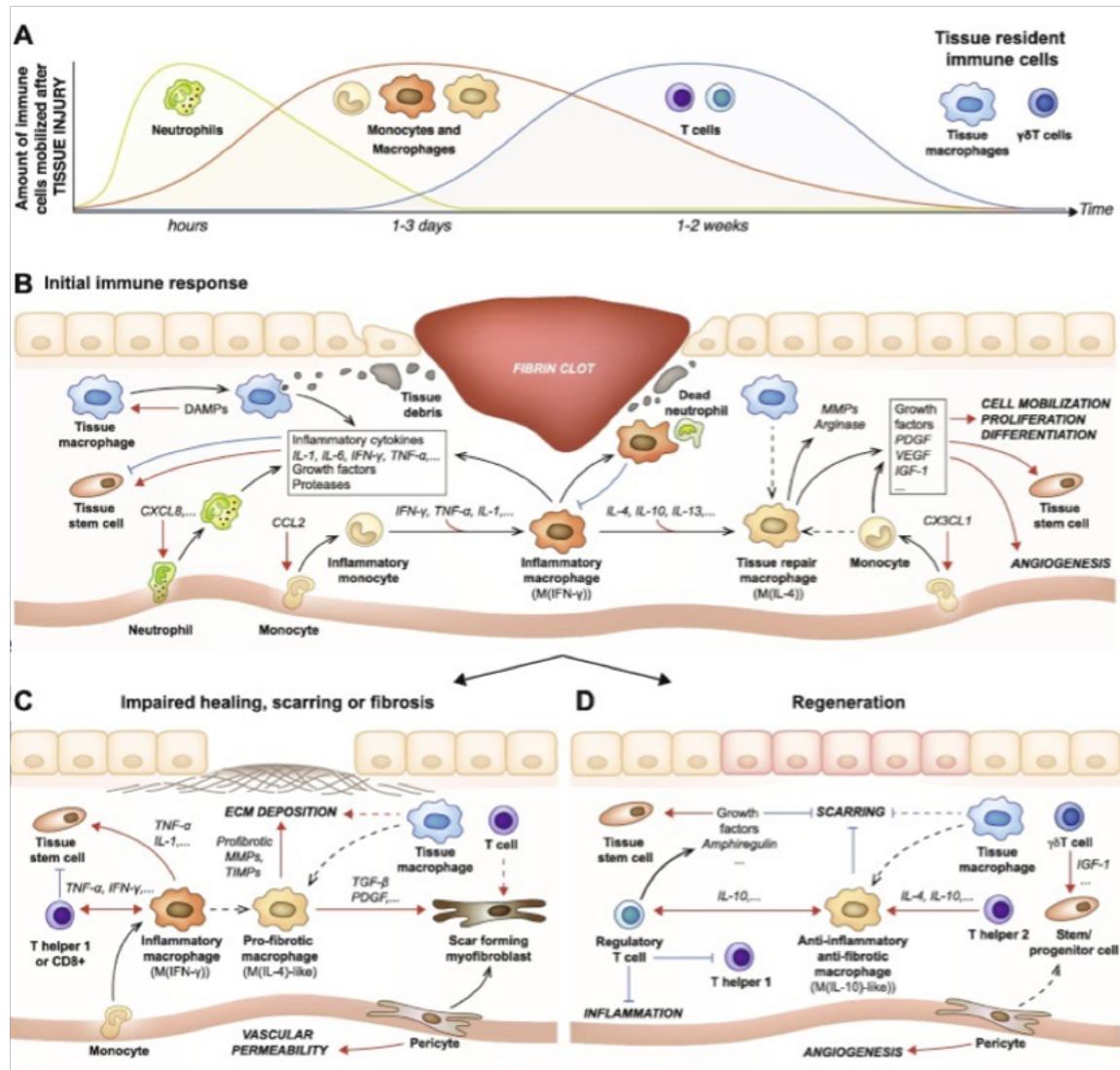


Figure 4.5

4.3. TISSUE HEALING

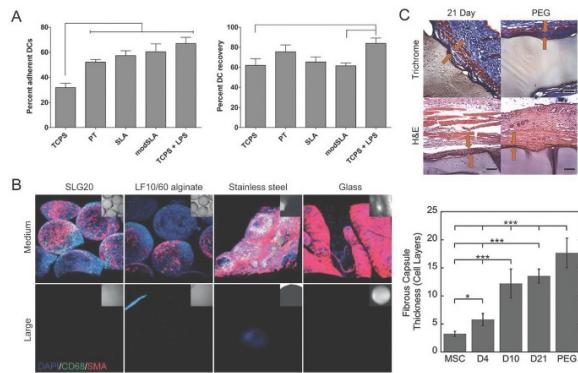


Figure 4.6

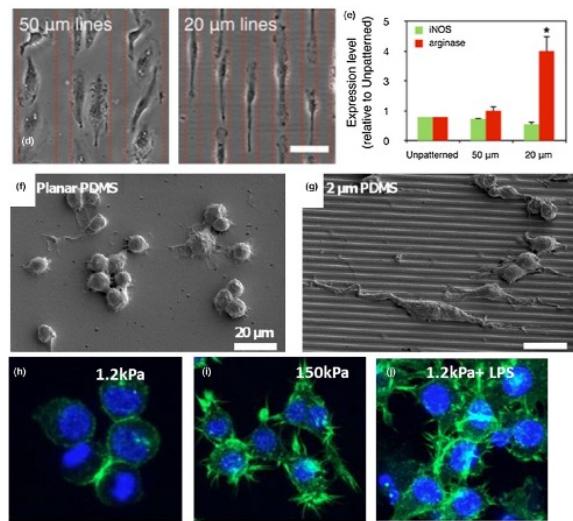


Figure 4.7: Macrophages and scaffold topography

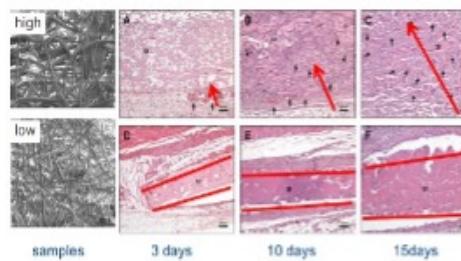


Figure 4.8

Chapter 5

Biorecognition

5.1 Molecular Biorecognition

Tissue engineering and regenerative medicine require an intimate understanding of the native ECM, together with the complexity of cell and tissue biology. For therapeutic applications, we should mimic the basic structure of ECM using a variety of synthetic or naturally derived materials and fabrication methods. The scaffold can be seen as a 3D growth environment:

- basic structural properties of ECM
- molecular cues to control bio responses
- protease sensitive site for enabling migration
- locally delivery of soluble factors for tissue remodelling stimulation

5.1.1 Functions of the ECM

- aids in locomotion
- transmits and distributes mechanical loads
- prevents premature mechanical failure
- partitions cells and tissues into functional units (scaffold architecture)
- acts as a scaffold that define tissue and organ architecture
- acts as a storage and dissipative devices for elastic energy
- acts as substrates for cell adhesion, growth and differentiation

These functions are defined by composition, structure, mechanical properties and repair response, which depend on tissue type, physiopathology, mechanical forces, damage and healing process.

Figure 5.1:the first example is a normal situation, good substrate. In an aged person we have a loss of elasticity. In the case of a wound, we observe a coagulation cascade and scar tissue formation - which is characterized by a compact tissue. The temporary material (crust) is used as scaffold.

5.1. MOLECULAR BIORECOGNITION

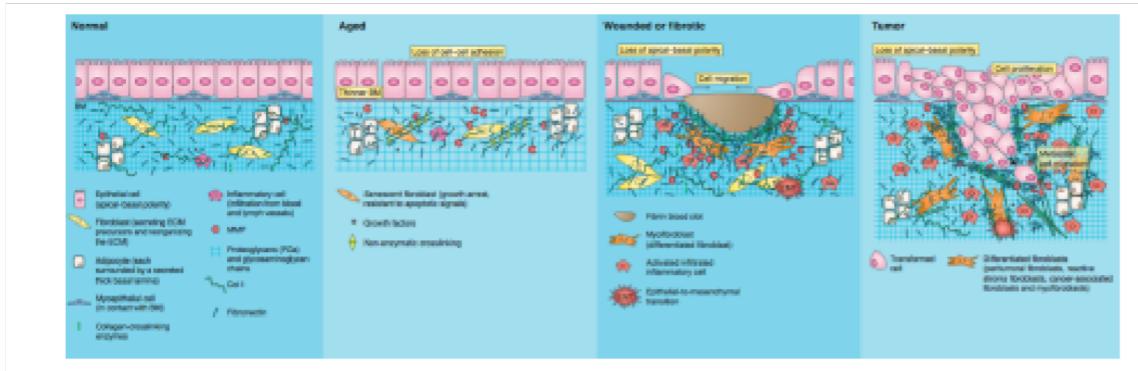


Figure 5.1: ECM functionalization

We can have healing by repair or regeneration. There is a higher rigidity in the case of temporary material. In the case of a tumour, we have an abnormal growth and migration inside - which they should not be present. The migration is caused by fibers not organising to close the hole.

5.1.2 Surface chemistry: biorecognition in ECM

Cell biology is governed by a complex series of interactions with the ECM. It is not enough to promote proliferation, we should control the adhesion involving the integrins (connected with few genes). We need a pool of molecules, since biological functions are orchestrated by a symphony of signals. Communication between cells and the cellular matrix is very relevant.

A **biomaterial** is a substance that has been engineered to take a form which, alone or as a part of a complex system, is used to direct, by control of interactions with components of living systems, the course of any therapeutic or diagnostic procedure in human or veterinary medicine. The aim is to recreate a basic environment before inducing regeneration. We need to upregulate proliferation at the beginning to create a population, then stop it for reaching ECM formation and achieving a therapeutic impact. Cells can interact with moving molecules through the gap junctions or through integrins; there are cells that must be very close e.g. skin, myocardium, in other cases integrins are mandatory. The ECM should also provide security for travelling molecules, as molecules should not be degraded and must be kept in the active conformation. Nutritional status is the bottle neck of tissue engineering, because in big wounds it is difficult to provide nutrition - as we need angiogenesis in order to have it. Hydrogels are cool because they can provide a lot of these requirements and can be combined to provide also mechanical support.

The cell language is based on a mapping (micro- and nano-patterning) of biopolymers (dynamic ECM) shapes to their complementary binding. Chemical process during cell activity:

- Chemical reactions (irreversible): provide free energy ($ATP \rightarrow ADP$)
 - Biopolymer (ECM) shape changes (reversible): provide control over chemical reactions

5.2. CELL INTERACTIONS

5.1.3 Biocompatible materials: foundation ideas

In order to design suitable materials, we should learn the biological pathways that lead to normal healing and reconstruction. Secondly, it is required to develop bio recognition surfaces that turn these pathways on and off. It is necessary to know specific affinities for the key molecules found in healing wounds that are associated with vascularised healing and regeneration (triggering local “unnatural” local healing). If biomolecules are immobilised at the surface, they must be in the correct orientation and conformation. Porosity should be engineered to induce vascularisation and less fibrotic tissue. Furthermore, the modulus matching of scaffold-biomaterial should match with their intended use (modulus mismatch should exacerbate the FBRx). Lastly, the engineered scaffold should be able to degrade into non-reactive substances at predetermined degradation rates to serve as a temporary guide for healing.

5.2 Cell interactions

Cells interact with the environment thanks to soluble factors, extracellular matrix and receptors. Signals form ECM and neighbouring cells:

- gene expression regulation leads to adult stem cell differentiation [into the lineage of interest e.g. osteoblasts in bone]
- tissue specific differentiation
- survival of primary cells
- Interaction with apoptosis: under external stimuli the cells may go to apoptosis. This may be useful in case of chemotherapy

The interaction between cells and matrix can be compared to a chemical reaction, where reagents can give rise to a reaction in a specific condition. Reactants are the cell + matrix, the specific condition is the biorecognition, we need integrins [without the plus (“+”) there is no reaction]. All the reactants must communicate with each other. Integrins are important for response]. In order to achieve regeneration, we need to activate a number of functionalities, which should be listed according to time. The scaffold is a reactant, it is added in the bioreaction. Of course we also have mechanical stresses involved in the reaction. We need to take into account the control system, which should not be downregulated e.g. activate tissue regrowth and activate control system to stop when the tissue volume is enough.

How does the body respond to the material? Cells are in suspension in the culture, which contains serum (proteins). The surface first absorbs a coating of ions and proteins/lipids, which are then recognized and bound by cells. Which kind of proteins will be absorbed by the implant? The quality of proteins depends on the chemical/mechanical properties of the scaffold and the context e.g. blood, skin, brain... For sure the first protein absorbed will be taken by plasma when we have bleeding. Depending on the proteins selected, the cells will adhere when the integrins of the cells can find some ligands among the proteins absorbed. We only have cell adhesion when we have biorecognition of the scaffold.

Surfaces must be designed in order to control protein absorption. Depending on the phenotype, we have different groups of integrins and also integrins specific for different genes. The aim of TE is to promote cell adhesion through specific integrins needed to achieve regeneration. Scaffolds and cells are not isolated, the empty spaces are filled by the ECM (providing proteins and water). This

5.2. CELL INTERACTIONS

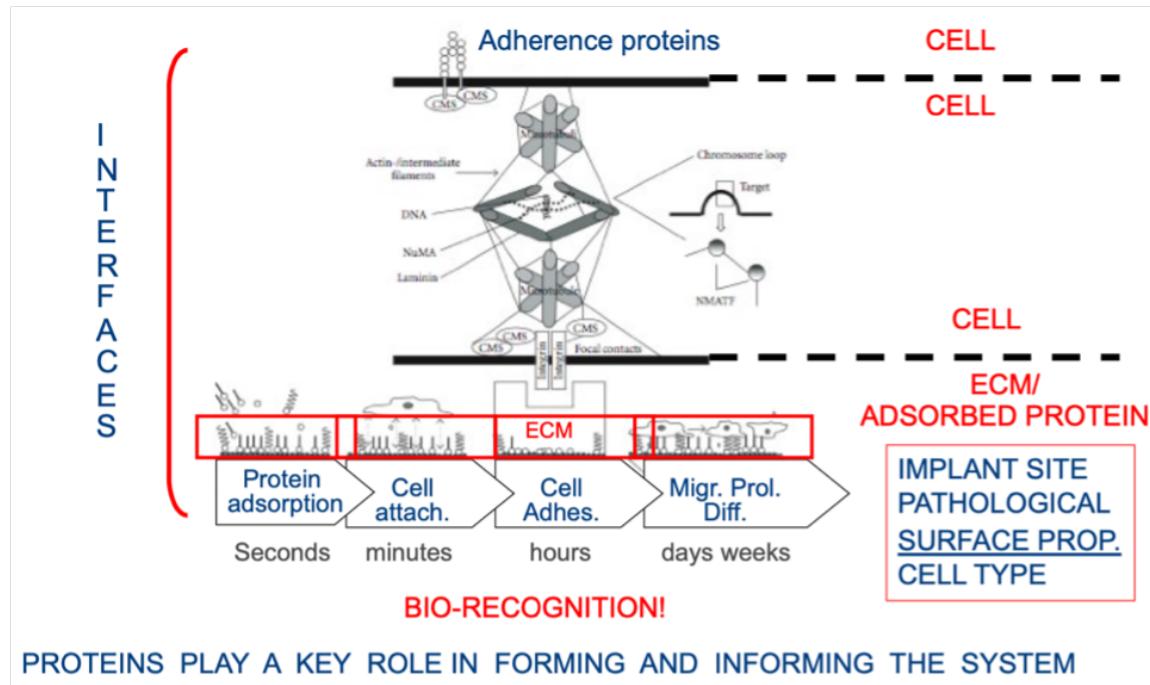


Figure 5.2

first step is crucial, depending on this we will reach failure or success. Therefore, the surface of the scaffold is really relevant. We can change the chemical properties, functionalize with new chemical groups, modify morphology to drive the interaction between scaffold and cells. The interaction between the implant and biological systems is a dynamic process. The biomaterial is characterized by specific properties. If we incubate the material in a single protein solution, where the protein is in active conformation. Depending on the surface properties, the biomaterial will absorb the proteins, which will remain active, or we will witness deactivation / degradation / modification.

1. Protein is absorbed with original conformation = still active
2. Denaturation during absorption = not active, switch off
3. Different conformation = different activity [dangerous, unexpected situation]
4. Degradation = no specific activity, negative response and inflammation [small pro-inflammatory peptides are released into the environment]

Having the protein active/inactive could be both positive or negative, it depends on what we need, what we want. The protein absorption mechanism is very dynamic, the absorbed proteins can also be released.

The protein substrate is important, we can change the material and see how it behaves. We could observe different interactions with the proteins according to the starting material. Biocompatibility is a two-way process, the host affects the implant and vice versa. Also depending on the domain we will have a different adhesion, we can find differential adhesion distribution.

5.2. CELL INTERACTIONS

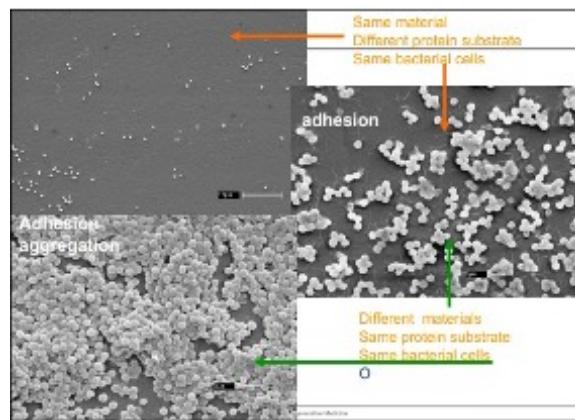


Figure 5.3

Figure 5.3: the aim was to produce a surface avoiding bacterial infection. The polymer is polyurethane, used for catheter production - antibacterial properties are really required. We see that the surface in contact with different protein substrates leads to different adhesion levels. When instead the protein substrate is equal with different materials, we see either adhesion or adhesion aggregation.

Figure 5.4: vascular graft that failed after 6 years of implantation. The implant failed because the

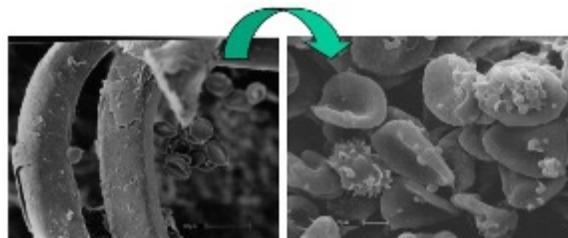


Figure 5.4

surface was completely destroyed. The bacterial cells (there was infection) adhered to the implant and this activated inflammatory response. Inflammatory cells digested the implant, made of a really strong polymer (same as parachutes). In addition, the implant started to detach in the circulation system + bacteria were able to attach to red blood cells. This is one of the few cases in which we do not want cell adhesion.

5.2.1 Molecular mechanism of cell adhesion

The orientation of ligands is critical for cell adhesion and biological function. The RGD sequence promotes cell adhesion and it is usually included in a long peptide – to avoid that RGD is involved in the link between cells, it would be invisible. Depending on the protocol parameters, we can achieve a nice or absent adhesion. The density is the same, the only difference is the orientation. For instance we could have a different density of signal, leading to a different outcome in adhesion. The density

5.2. CELL INTERACTIONS

of signal is important for the function. Our aim is to reach an equilibrium, not too much or too little adhesion. In the second scenario the cell is able to move, good choice if cells are required to migrate into the scaffold. If instead we need a fast formation of a layer we can choose the first case. No absorption = no adhesion at all. This may be useful in some cases, e.g. in blood vessels to prevent thrombus, release of proteins from nanoparticles in cancer treatment. Cell adhesion is mainly controlled by the surface. How can we know whether the conformation and orientation is correct? We can employ cell sensors, screening.

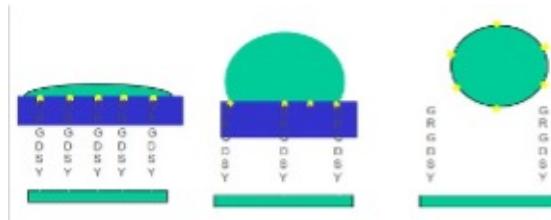


Figure 5.5

5.2.2 Cell adhesion

Cell adhesion is a tightly regulated and dynamic biological process. It is central to physiological and pathological processes and critical to biomedical and biotechnological applications. Adhesive interactions involve:

- anchorage (promotes migration, tissue organisation)
- signaling (promotes activation, survival, proliferation, differentiation)

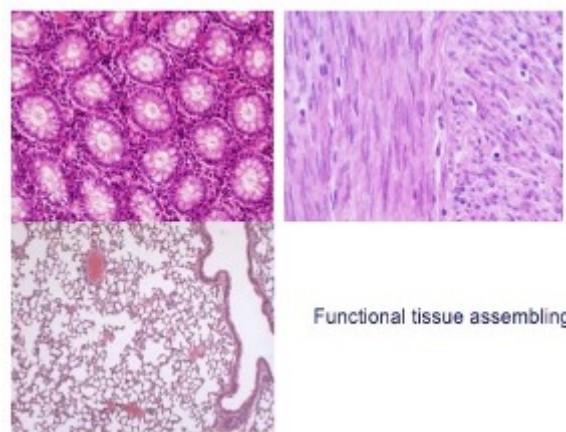


Figure 5.6

In figure 5.6 starting from top left we see the functional tissue assembling of glands (radial section), muscles, and lung/alveolar tissue. In the case of muscles, functional tissue should support mechanical stress. In glands we have cells forming very well polarized tubes for transport, which

5.2. CELL INTERACTIONS

are called capillary tubes. For lung tissue we have blood vessels, we need to have a small surface and permeability for gas exchange. The air should move into the bloodstream, so the tissue should be very thin and adherent to the vessels. We have a low amount of ECM, the mechanical support is provided by cells themselves. The structure is function-dependent. Assembly is driven by ECM and adhesion patterning. The morphology should depend on the function. Thin layer of ECM in the basal lamina, collagen fibrils. This provides elasticity, the fibrils are not a bundle, but organise somehow.

5.2.3 Cell-material interactions

Cells cannot adhere to synthetic surfaces, as there is no biorecognition (ECM performs better). Cell adhesion to synthetic and bio surfaces occurs through a specific receptor interaction with adhesion protein/motifs:

- proteins adsorbed from physiological fluids (fibronectin, vitronectin, fibrogen)
- ECM components present or deposited by cells (fibronectin, collagen, laminin)
- biospecific sequences engineered on surfaces (RGC,YIGSR) for biorecognition

Adhesion receptor families are cadherins, selectins, HSPG, integrins, Ig superfamily. Specific integrins act on specific receptors, so we must be precise. For instance, alpha5beta3 can recognize the ligands into fibronectin, collagen is recognized by alpha2beta1.

How to measure the adhesion strength of the cells? Experiment by Gallant and Garcia (2003). They provided two samples, performed centrifugation and measured cells attached and cells not adhered. In this way you can measure the difference in the strength of the adhesion.

Depending on the surface chemistry, we will have a different cell adhesion rate. According to the protein coating, e.g. either fibronectin or type I collagen, we will have different biological performance. A number of evaluation methods can give us an idea of the adherence strength.

Figure 5.7 design scaffold for neuron regeneration. Adhesion is necessary, but neurons also need to form connections with other neurons. RGD goal: neurite outgrowth, strategy: RGD-functionalization. They provided the material with fibrin, fibrin with low RGD and fibrin with high RGD. RGD is the classical adhesion peptide. We observe a different response: high density is too much, instead with a low amount of RGD we have a good outcome with a huge number of connections. When we increase RGD, the surface becomes more adhesive; this is mainly due to the fact that the cells are too sticky, but also because we must reproduce a biological environment. The amount of RGD should resemble the natural one. In fibronectin, the RGD content is 1 per chain, low amount. We have to move inside the natural range of pH, etc, ... We want natural conditions. The language of the cells cannot be changed, we have to adapt our language. Signal is the combination of the ligand and density.

Adhesion is completely different in 2D and 3D. Cell adhesion is characterized by three stages: attachment cell body, flattening and spreading, and organization of the actin skeleton with the formation of focal adhesion between cell and its substrate. The strength of adhesion becomes stronger with the length of time a cell is allowed to adhere to a substrate or another cell.

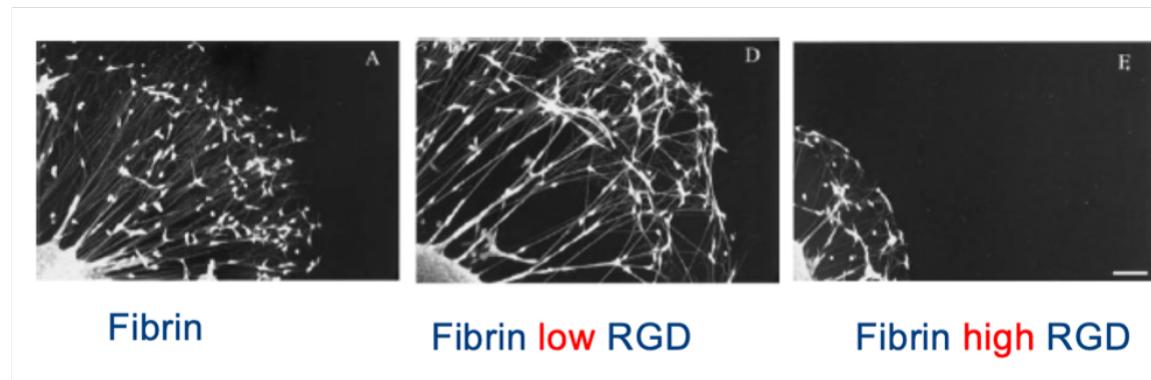


Figure 5.7

5.2.4 Cell organization

The cell's adhesive interactions with the surrounding ECM (number of adhesive motifs, distribution, density) and neighbouring cells define cell shape and organisation, controlling functionality. This environment regulates the cell survival, differentiation, proliferation and migration.

Chondrocytes are exposed to compressive forces, interstitial fluid flow and adhesive cues (cytokines) for cartilage maintenance. Chondrocytes are in a lacuna, they should behave like pillars and maintain their shape. Soluble and matrix-bound GFs and flow induced mechanical forces on blood vessel wall, endos after polarity, cell-cell contacts, and degrade the surrounding basement membrane and stromal ECM in order to migrate and form tubular sprouts. Adhesive and mechanical cues drive cell organization. Misregulation of the mechanism induces mechanical and structural changes in the ECM, and transformed epithelial cells migrate towards vasculature and eventually metastasize.

ECM-dependent regulators can be associated with 2D, 1D and 3D migration. In turns influence intracellular pathways that govern the migratory phenotype. 3D are characterized by pore size and interconnection (cross linking degree). Aligned fibers are randomly distributed with low density in the scaffold.

Adhesion and migration are controlled by the ECM composition, stiffness of the material and ligand density. While using fibers, something changes: by having a new parameter, aligned topography, we obtain different architectures. In the case of a mixed fibrous scaffold, we also have aligned/random, elastic behaviour, cross linking. Sending seed cells on the different structure, we will see diverse behaviour for orientation, migration, etc. Architecture can play a huge role.

The substrate contractility regulates 3D migration, regardless of pore size. Stiff fibers vs soft fibers: when the cell lands on stiff fiber it can adhere, but becomes very stable and not able to modify the body/migrate. Instead cells on soft fibers are able to move and form protrusions. Since cells in nature are connected to ECM fibers, when they move the ECM will also follow the contraction of the cell body. If the cell adheres to the soft fiber, the same natural movement can occur. When the substrate is too rigid contraction cannot occur, healing will be different. The scaffold should be soft enough to follow the reorganization needed by the cells.

5.2. CELL INTERACTIONS

Figure 5.8: two scaffolds prepared with the same polymer, the difference is in pore size and

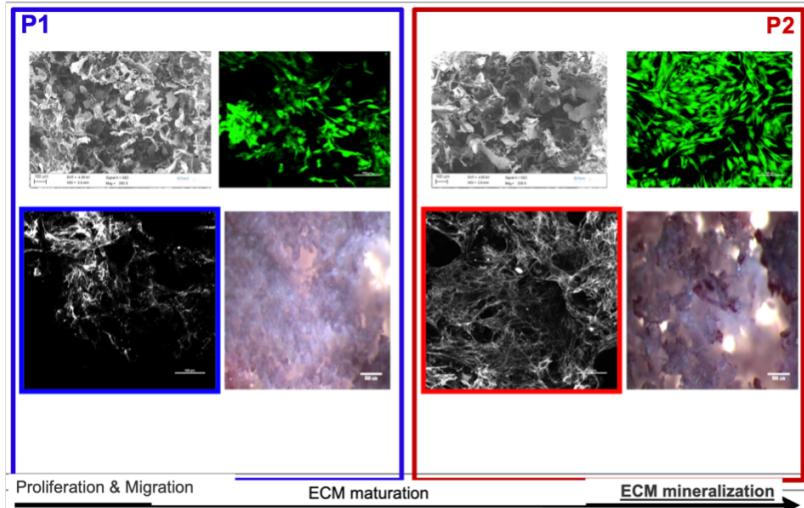


Figure 5.8

distribution. The scaffolds are built for bone regeneration; we should take into account the presence of collagen fiber in a continuous network. Osteoblasts should infiltrate the scaffold for building the network, porosity should allow this. Once the network is formed, we have the deposition of the hydroxyapatite on top of collagen fibers. Difference between P1 and P2 in terms of cell distribution: in P2 uniform, in P1 clusters (interconnection not complete). When we observe clusters of cells, we expect that the osteoblasts are not able to build a continuous network, due to tight collagen + no mineralization. When cells are not able to migrate, mineralization cannot occur. A continuous network is also required for capillary formation. The parameters can be fully controlled with scaffold design. In order to improve the performance of P2 scaffold we could functionalize it with collagen or drug release systems.

The scaffold surface interacts with the cell through the ECM (biorecognition). Our aim is to reproduce in vitro a working environment, working for the cell and solving a specific task. Depending on the context, we can have the bioreaction: the cell recognizes the surface of the scaffold if it is functionalized or for protein absorption. We have to refer to the cell population we are interested in, we have specific integrins.

5.2.5 Methods for modulating receptor-ligand interactions

1. Natural ECM biomaterials: biologically relevant environment, but poor mechanical properties and inconsistent reproducibility
2. Whole ECM adsorption
3. Synthetic linear binding motif: surface functionalization. We need to define the density of the signal, the protein of interest, the stability (quantify the time for which the signal should remain, orientation, homogeneous or pattern distribution).
4. Spatially oriented binding motif

5.3. BIORECOGNITION REQUIREMENTS

5. Nanopatterning with nanolithography: mix different molecules and patterns, as well as technologies
6. ECM-like biomaterials

5.3 Biorecognition requirements

The biorecognition process will have a good outcome only if the absorbed proteins are the right ones according to the following parameters:

- The absorbed proteins must contain sequences (ligands) recognizable by dedicated cellular receptors: among the first we remember RGD, YIGSR, IKVAV and RETTAWA, among the second we cite integrins. Integrins are heteromeric transmembrane receptors that mediate cell - ECM interaction with different glycoproteins (among whom fibronectin and vitronectin) and collagen fibers.
- The ligands mentioned above must be oriented outward, they must not be hidden or used by the scaffold to link with the proteins. If this happens, the sequences won't be able to be recognized by the cellular receptors.
- The absorbed proteins must have a stable structure: they must not denature, take on conformations with unexpected functions (es: prions!) or hold native functions that could hinder the regenerative process (es: trigger clotting formation) or release peptides (that could be pro - inflammatory, leading to chronic inflammation).
- The array of absorbed protein must expose the right number of ligands, with the right density. In particular for the regenerative aim, we want our ligands to be in the right number and density to allow for macrophage adhesion while still allowing them to move. We also want them to stretch across the binding sites, since this stretching has been correlated with the polarization towards the M2 phenotype.

Chapter 6

Hydrogels

Hydrogels were the first biomaterials rationally designed for human use, especially for soft tissue engineering. They are a particular class of materials, exhibiting both solid-like and liquid-like properties, consisting of hydrophilic polymer chains and water, that occupies the interstitial spaces (pores) that are defined in the 3D network constituted by the cross-linked polymeric chains.

Water is the major constituent of hydrogels and could reach more than 90 percent of the weight of the material - although its quantity is dependent on the degree of hydrophilicity of the polymer chains. Cells can be added inside the hydrogel before or after gelation. In the first case, one must make sure that the gelation itself won't damage the cells. In the second case, porosity must allow for uniform cellular colonization.

Hydrogels' characteristics can be modulated according to three main variables:

- Chain composition: they can be natural (usually polysaccharides) or synthetic polymers. The main subvariables are chain length, degree of hydrophilicity and presence of ligands recognizable by cellular receptors (hydrogels must be biorecognizable). The utility of synthetic polymers is that they can be functionalized with these kinds of ligands to allow cell adhesion, but degradability will become a problem.
- Cross-linking nature: strategy employed to connect the polymeric chains. This act is also called gelation. This could be:
 - Chemical crosslinking: the polymer chains are covalently linked. This linkage can be obtained in an enzymatic way (if enzymes capable of interacting with the chosen polymers exist). The enzymatic crosslinking is also useful because it guarantees the degradability of the scaffold via host enzymes or in a non - enzymatic way, exploiting specific reactions that may require specific reagents.
 - Physical crosslinking: the polymer chains are held together by molecular entanglements (temporary spatial constraints) and/or secondary forces such as ionic, hydrophobic, and hydrogen bonds. Chemically crosslinked hydrogels always present some degree of physical interaction as well, but the contrary, of course, does not occur. Physical crosslinking can be obtained via thermal treatment, pH changes and treatment with organic solvents.
- Network nature: it's the overall 3D structure defined by the crosslinked polymers. Its main characteristic is the number and the dimension of the interstitial spaces (pores), influenced

by polymer chains length and by the number of crosslinkings. Pores number and dimension, together with the degree of hydrophilicity of the polymer chains, will determine the maximum amount of water hosted by the hydrogel. This in turn will impact the swelling capacity, one of the main characteristics of these kinds of scaffolds.

Hydrogels can also be designed with the only goal of carrying modded cells to a particular region of the body: in that case non biocompatible, non biorecognizable synthetic polymers can be used to avoid premature hydrogel degradation and unwanted interaction with the cargo and to carry the cells to the targeted location.

Hydrogels can also be used to design drug releasing systems: they can function as surrogate matrices to carry modified cells producing a specific therapeutic agent. In order to protect these cells from the host's immune system, the hydrogel itself must be coated by a semipermeable membrane, that must allow the entrance of nutrients and growth factors in the hydrogel and the exiting of the therapeutic products.

Chapter 7

Fetal healing

7.1 Scarring

A scar is a densely packed disorganized collagen bundle, with absence of hair follicles, sebaceous glands and other appendages. Scarring and fibrosis dominate some diseases in every branch of medicine and surgery. Examples: skin incisions heal with scars (pathological processes as keloids, hypertrophic scars, ...).

7.1.1 Adult scarring

The mechanism of scar formation involves inflammation, fibroplasia, formation of granulation tissue, and scar maturation. The acute inflammatory response (pro inflammatory mediators) is followed by the proliferation of fibroblasts, which are cells responsible for synthesizing various tissue components, including collagen and fibrin. During the acute inflammatory phase, circulating progenitor cells migrate to injured tissue. Rapid cellular proliferation occurs, which ultimately results in the formation of new blood vessels and epithelium. Fibroblasts then differentiate into myofibroblasts, which are the cells responsible for collagen deposition and wound contraction. Scar formation ultimately results from excess accumulation of an unorganized extracellular matrix. Although scar remodeling occurs for months to years after the initial injury, complete restoration of the normal extracellular matrix architecture is never achieved.

7.2 Embryonic healing

In contrast to adult wounds, early gestation fetal skin wounds repair rapidly and in the absence of scar formation. The fetus is able to regenerate tissues by assembling collagen fiber in a well organised structure. The biology of fetal repair must be understood; in particular, cellular and matrix events may provide insights to help to modulate adult wound repair to become more fetal-like.

7.2.1 Fetal extracellular matrix

We should try to mimic the embryonic development procedure, where the ECM is elaborated in parallel with cell differentiation and growth. In particular, our aim is not to obtain a normal ECM analogue (mature scaffold), but a wound bed matrix analogue suitable for regeneration. Collagen represents a "mature" scaffold, forming a microenvironment suitable for fully differentiated cells.

7.2. EMBRYONIC HEALING

If we build a scaffold based on collagen type I, cells which don't usually reside in an ECM made extensively of collagen type I will not interact with the surface, leading to a pathologic response. Example: chondrocytes usually interact with collagen type II, so if collagen type I is present cells will form fibrocartilage.

The fetal extracellular matrix is optimized to facilitate cellular migration and proliferation, which may have important implications in wound healing.

- collagen content: high content of collagen type II and V, increased collagen synthesis.
- hyaluronic acid: in scarless fetal wounds, the hyaluronic acid content of the extracellular matrix is increased more rapidly than in adult wounds, major component.
- adhesion proteins: scarless fetal wounds have an enhanced ability to up-regulate extracellular matrix adhesion proteins, such as tenascin and fibronectin
- ECM modulators: fibromodulin inactivates transforming growth factor (TGF)-beta, a key cytokine involved in wound healing, and has been shown to have an antiscarring effect during wound repair.
- non-sulfonated GAGs

7.2.2 Fetal mediators of scarless repair

- lack of inflammation: decreased platelet degranulation and aggregation, anti-inflammatory cytokines
- fibroblast cells with high migration rate (affecting collagen depth and cross linking)
- increased number of HA receptors
- myofibroblasts appear earlier and then disappear

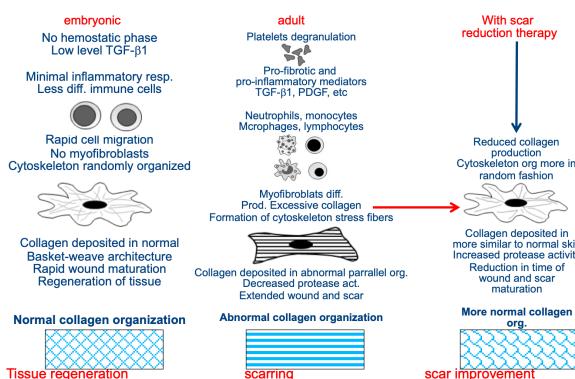


Figure 7.1: A lesson from fetal healing: scarless regeneration

7.3. HYALURONAN

7.3 Hyaluronan

Hyaluronan is a hydrated gelatinous material synthesized from basal side of an epithelial sheet. It creates a cell-free space into which cells can proliferate and migrate, perform the diffusion of nutrients, metabolites,... It is able to resist compressive forces and acts as space filling material in embryogenesis. It is degraded by hyaluronidase.

Hyaluronian is composed by the simplest GAG with a very long chain length (25000 disaccharide units. It lacks sulfated sugars and is usually not attached to protein (used as filler). The high negative charges on the glucuronic acid attract Na^+ ions and water.

7.4 Biology of regeneration

Regeneration of amputated limb can occur at an adult stage in amphibians and fish. This complex process is enabled by specific tissue regeneration mechanisms. For instance, in adults of *Homo Sapiens* the liver regenerates spontaneously. Furthermore, we have little capacity to regenerate tendons and ligaments.

Chapter 8

Scaffold design

The dynamics of regeneration vary from tissue to tissue according to the hierarchy of tissue or organ function. Remember that “the ability of a material to perform with an appropriate host response in a specific application”. Important consideration for the materials:

- sourcing of functional cells: if the scaffold requires a pre-loading of cells, we need to discuss which kind of cells to employ beforehand e.g. stem cells, cells from the patient (bone marrow, amniotic liquid, etc)
- GF regulatory systems: synthetic polymers with no biorecognition properties require functionalization
- immuno acceptance: e.g. force stem cells to become osteoblasts or modulate the immune response in order to shorten the inflammation and boost regeneration

Biological information is required for performing in vitro tests and for functionalization (linked to specific biological pathways, essential to thoroughly describe the mechanisms).

Scaffolds can be grouped in two categories:

- conductive: provide and maintain a 3D environment that supports a passive cell infiltration, creating a pseudo micro environment. The limitation is that they are not able to provide enough info to promote full regeneration in most applications.
- inductive: designed to closely mimic the native cellular environment and may contain bioactive molecules and naturally or synthetic analogues of structural, functional or specialised proteins and proteoglycans. They can increase the biocomplexity of the system.

A scaffold is part of a complex system. It is used to direct, by control of interactions with components of living systems, the course of any therapeutic or diagnostic procedure. A scaffold is composed by different materials e.g. natural and synthetic polymers. The geometry should be suitable for the specific application e.g. allow migration, cell distribution in 3D in functional manner, alignment. Polymers can be combined with other materials or drugs. If the interaction and biocompatibility are present, we will have a suitable response.

8.0.1 Tissue engineering strategies

- "just" scaffold *in vivo*: the body helps with the regeneration, if possible it's the best strategy we can follow.
- scaffold + cells implantation: we need to define why we wish to culture *in vivo* e.g. ECM production, differentiation,...
- cell sheet engineering: fabrication with cells (stem cells) from the patient.

Cell sheet engineering is the only bottom-up approach, starting from the material and not the biology.

8.0.2 Biocompatibility requirements

Revised performance criteria for the 4th generation of biomaterials:

- tailored biodegradation
- amenability to engineering design and manufacturing
- induces cell and tissue integration
- smart (i.e. physiologically responsive)
- instructional (i.e. controls cell fate)
- mechanical strength and function (i.e. mechanical signalling)

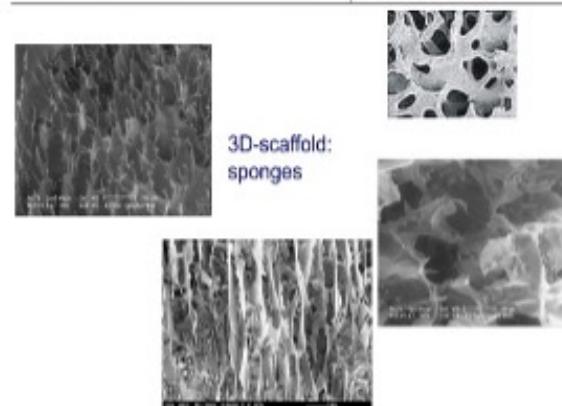


Figure 8.1

Fig 8.1 repair of trabecular bone with 3D-scaffold sponges. All of them are artificial scaffolds, while on the top right we have the natural sponge. The porosity can be oriented or random, with different geometries. The scaffold should promote adhesion, proliferation and migration. Hypoxia should be avoided, maybe with early angiogenesis. *In vivo*: in the case of bones, we have osteoblasts adhesion and migration, ECM production. In order to avoid hypoxia and necrotic tissue formation it is required to achieve vascularization, we need to supply oxygen to stimulate angiogenesis. *In vitro*: millimetres not microns, the problem is more tough. We could culture in a bioreactor e.g. chamber with medium forced to go through, or include into the scaffold oxygen donors.

8.0.3 Experimental data discussion

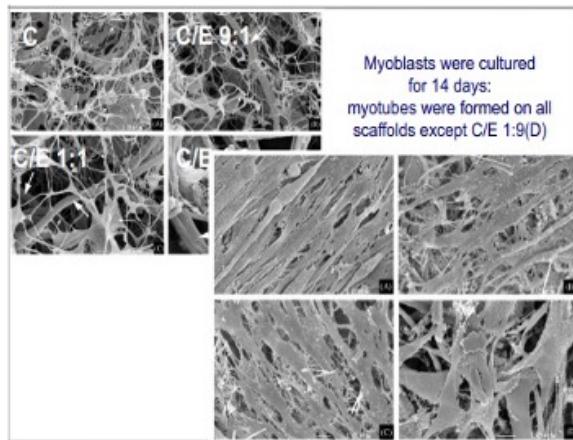


Figure 8.2

Fig 8.2: muscle regeneration. Use collagen, elastin and glycosaminoglycans. GAGs control water content and mechanical properties. Elastin is required for muscle elasticity, collagen for the strength. It is necessary to find the optimal ratio among the components. By changing the ratio, we have 4 different architectures:

1. C: sponge, similar to natural behaviour of collagen
2. C/E 9:1: the big fiber starts to appear (elastin)
3. C/E 1:1: similar content
4. C/E 1:9: a lot of big fibers pink fiber: elastin fiber

C might not be optimal, as well as C/E 1:9, as they are quite different from biological setting. In vitro test: after two weeks myoblasts were cultured for 14 days and myotubes were formed on all scaffolds except C/E 1:9(D). We witness a different organisation of the myotubes; in the last example, myoblasts are not able to organise as the condition is very far from physiology. By considering the test, the best samples seem to be C/E 9:1 and C/E 1:1 .

Tissue engineering is 3D assembly over time of vital tissues-organs by a process involving cells, signals, and the extracellular matrix. In this case we have a scaffold for bone regeneration and osteoblast proliferation (fig 8.3). The cells (green) are able to adhere, proliferate and migrate. The scaffold is then completely covered in cells. We then witness tissue-specific ECM production, mineralization. This is a great starting point, but we have to better understand the degradation process of the scaffold and also vascularisation! How can we check in vivo vascularisation? We need to perform co-culture of osteoblasts (angiogenesis factors + collagenic structure for capillary network formation) and endothelial cells (use empty spaces to assemble as a tube). To obtain a physiological situation we need a careful design.

8.1. SCAFFOLD EXAMPLES

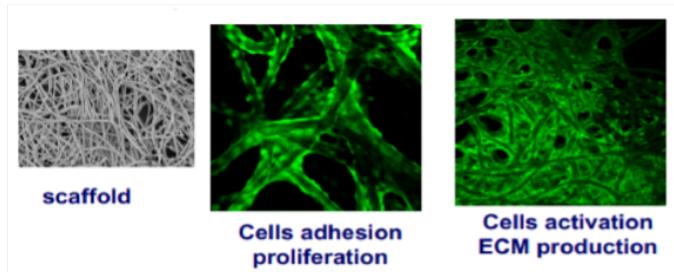


Figure 8.3

8.1 Scaffold examples

8.1.1 Example 1

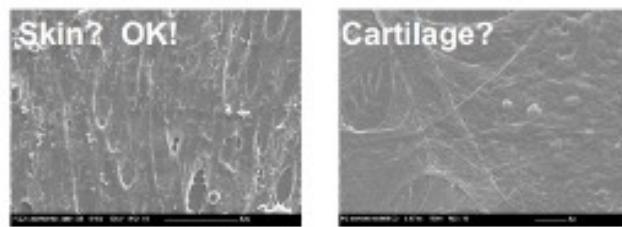


Figure 8.4

Fig 8.4: film with same polymers, same architecture, different cells: keratinocytes and chondrocytes. The two films were completely spread. Cells grown in a flat dish tend to behave as individual cells or forming a monolayer, whereas cells cultured in a 3D space are more likely to assume the characteristics of a particular tissue. Cartilage, once grown flat, is almost impossible to shape into joints. The scaffold here, in the case of cartilage, is inducing the loss of the original phenotype, so it is not biocompatible. In the case of the skin, it is promising for biocompatibility, very compact and well connected monolayers.

8.1.2 Example 2

Fig 8.5: in this case we do not have biocompatibility at all. The cells adhere to the polymer but do not migrate, so we should increase the porosity. After few days: we have a multi-layer as result, no migration and no biocompatibility.

8.1.3 Example 3

Fig 8.6 suine primary urothelial cells, 6 days after seeding. The right image has the geometrical shape of the cells, the phenotype is ok. Left: cells don't connect and are disorganized, so they do not communicate. They are more round shaped, so the adherence is not good, there is no activation.

8.1. SCAFFOLD EXAMPLES

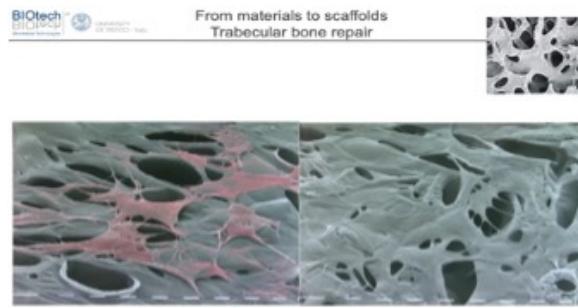


Figure 8.5

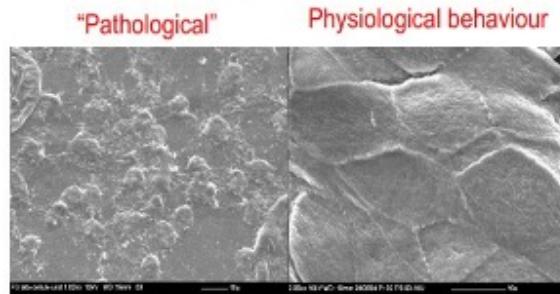


Figure 8.6

8.1.4 Example 4

Fibroin micronet (Fig 9.5) human microcapillary endothelial cells (HDMEC) and primary human osteoblast cells (HOS) in coculture (10 days). Very good results, empty tubes, but it is promising. They can be put into the scaffold and then maybe used as vessels. Issue: We have to pay attention if the vessels connect to our circulatory system!

8.1.5 Example 5

Figure 8.8: pre-vascularized fibroin net, in vivo anastomosis with host vasculature. Brown tubes: in vivo, blood cells are present and anastomise. Left: loaded with osteoblast, let grow and then implanted. Fast angiogenesis, not too many vessels, only in the surface Right: loaded with osteoblast, immediately implanted. In vitro human cells and implanted in rat to identify the difference. On the right there are some capillaries that were produced and there is not red blood cells.

8.1.6 Example 6

Figure 8.9: from a NaCl solution salt crystals will form a sponge. The porosity depends on crystal size. We can then apply gamma rays treatment and add either silk fibroin or P(d,l)LA.

MISSING LECTURE

8.1. SCAFFOLD EXAMPLES

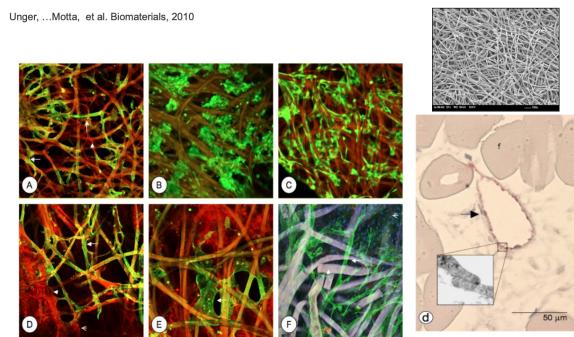


Figure 8.7

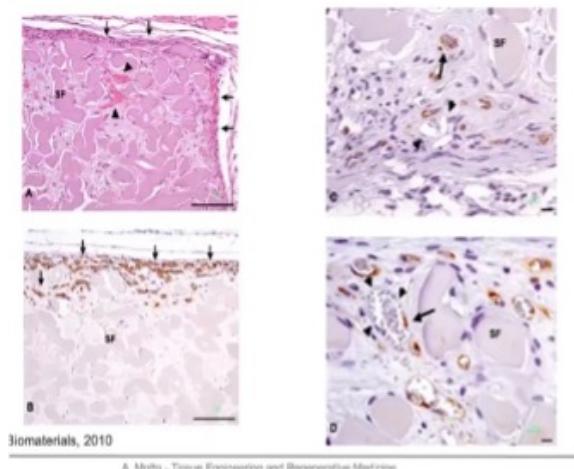


Figure 8.8

8.1.7 Ectopic implant in rat

An ectopic implant is done in a site that is not natural, in the studied case under the skin for bone.

8.1. SCAFFOLD EXAMPLES

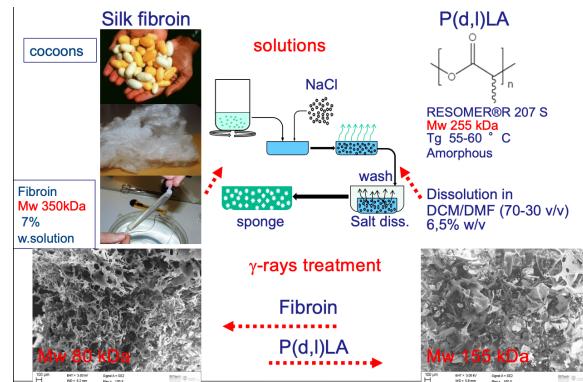


Figure 8.9

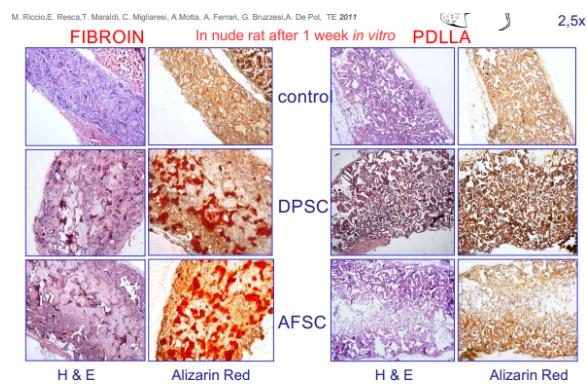


Figure 8.10: **1. Fibroin.** In the **control** there is no red sign at all so for this reason we can say that there is not induction. There are cells (blue) but they are not infiltrated in the scaffold. There are no sign of mineralization. No osteoinduction. In **DPSC** and **AFSC** there are red nodules and that means that it becomes osteoinductive and force stem cells to become osteoblast and also mineralization. In the **H & E** column we can notice a regeneration framework. **2. PDLLA** In the **control** there is not red sign so there is no induction. There are cells (blue) but are not filtered in the scaffold. There is no mineralization or osteoinduction. In **DPSC** and **AFSC**, even in the case of pre-loaded cells there are not red signs and also there is not osteoinduction. In the center there are not cells and that suggests us that the cells started to go into apoptosis.

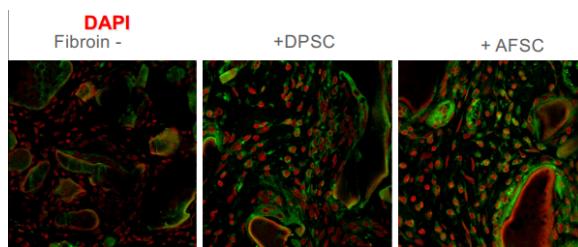


Figure 8.11: In order to assess which kind of cell is working, pre-loaded cells or osteo cells, we can do an anti-human staining.

8.2. CELL SHEET ENGINEERING

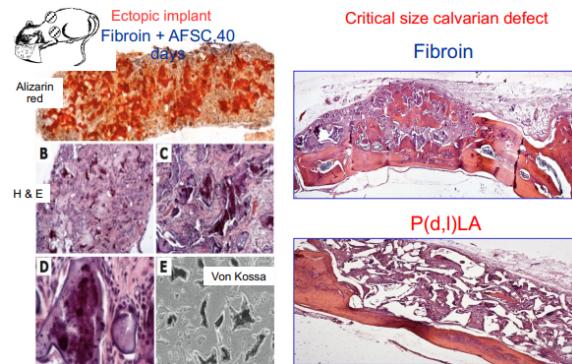


Figure 8.12: Intaosseous implants. The fibroin enhance the new bone formation and supports regeneration. In P(d,l)LA there are lots of holes and the regeneration is not completed. It is possible to evaluate also the quality of the bone and we can see in the fibroin it was good. There is a mixture of human and murine cells that are working together.

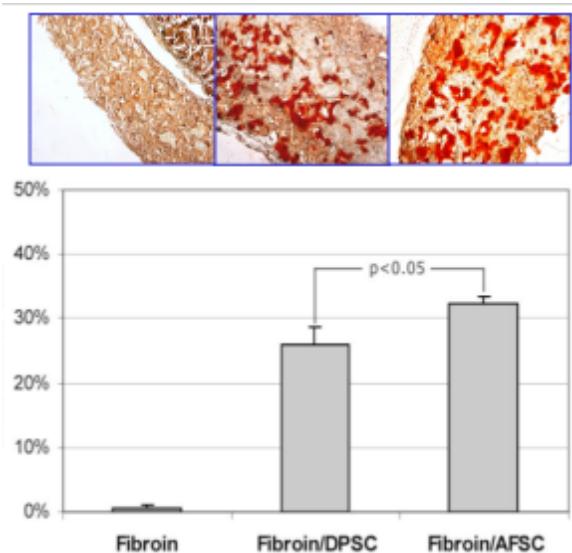


Figure 8.13: Areas of mineralization. In the figure it is possible to see the areas of mineralization on fibroin scaffolds after 4 weeks implantation. It was calculated on 5 transversal sections cut at interval of 1 mm in the mid region of different fibroin stained by Alizarin Red. There is no mineralization in the scaffold with only fibroin, while there is mineralization in the two scaffold that present also AFSC and DPSC. There is no mineralization in the three scaffolds with PdILA.

8.2 Cell sheet engineering

Cells sheet engineering is based on thermo-responsive polymers. They can be used in different sites like oral mucosa that present a very difficult regeneration, in periodontal ligament regeneration, but also in myocardium regeneration (as an alternative to heart transplantation). By transplanting

8.2. CELL SHEET ENGINEERING

single cell sheets directly to host tissues, skin, cornea, periodontal ligament, and bladder can be reconstructed. Additionally, the creation of co-cultured cell sheets from dishes with dual temperature-responsive domains, also allows for the re-creation of higher-order structures such as the kidney and liver.

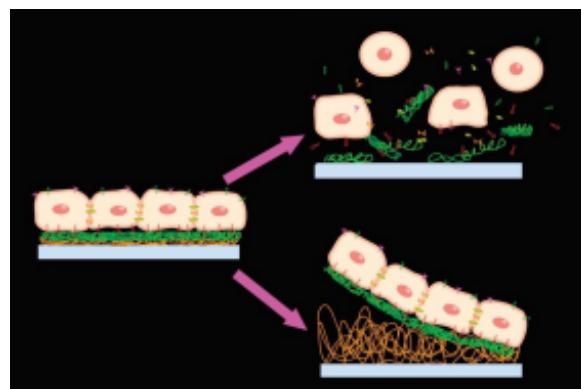


Figure 8.14: Cell sheet harvest deposited ECM (green), as well as membrane proteins, so that confluent, monolayer cells are harvested as single cells (upper right). The temperature-responsive polymer (orange) covalently immobilized on the dish surface hydrates when the temperature is reduced, decreasing the interaction with deposited ECM. All the cells connected via cell-cell junction proteins are harvested as a single, contiguous cell sheet without the need for proteolytic enzymes.

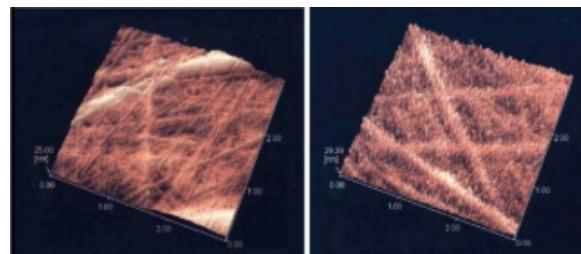


Figure 8.15: Atomic force microscope images of temperature-responsive culture dish surfaces. Nongrafted, polystyrene culture dish surfaces (left) and poly(*N*-isopropylacrylamide)-grafted culture dish surfaces (right) were examined in air

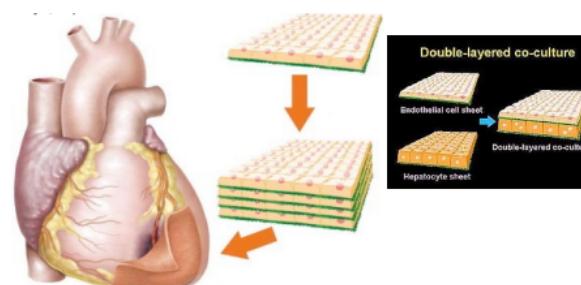


Figure 8.16: Cardiac Tissue Reconstruction Based on Cell Sheet Engineering

8.2. CELL SHEET ENGINEERING

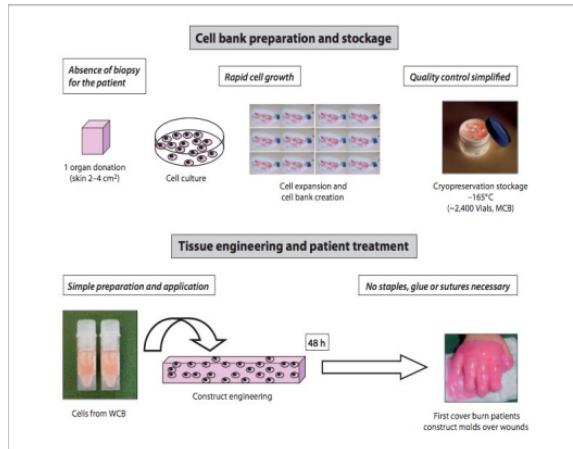
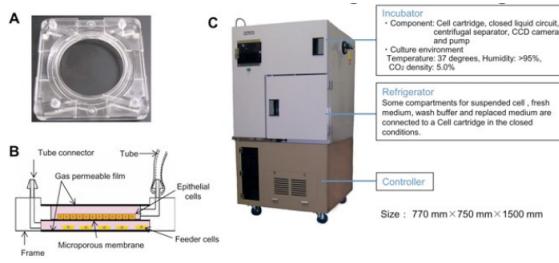


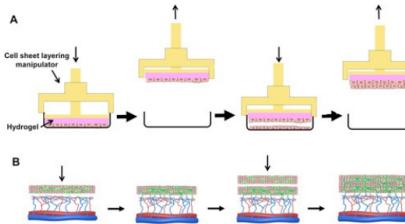
Figure 8.17: Burn skin

8.2.1 Current challenges and strategies



Schematic of the manufacturing cycle to supply patients with autologous cell sheets.

Every clinical-grade cell sheet is fabricated through the following scheme: cell isolation from patient's tiny amount of tissue; primary and/or expansion culture; cell seeding on temperature-responsive culture vessel where a cell sheet is formed.



Unique technologies for 3D tissue fabrication by layering cell sheets. (A) Stamping technology to stack cell sheets in layers with the manipulator. (B) Vascular network formation inside of the stacked cell sheets.



Prototype of T-Factory and flexible modular platform. Various types of compact isolator are currently under development: (A) module for cell sheet layering; (B) cell seeding / medium replacement; (C) CO₂ incubation; (D) the central module that transfers cell sheet intermediates between modules during the manufacture of cell sheets; and (E) the isolator where starting materials are manually processed before subjected to an automatic manufacture.

Figure 8.18: Current challenges and strategies in cells sheet engineering.

8.3. CELL ENCAPSULATION

8.3 Cell encapsulation

The cell encapsulation methods consists in the entrapment of cells in microcapsules or microbeads starting from a suspension of cells in polymeric solution that can be solidified by chemical or physical methods. During the encapsulation the polymer cross-links by chemicals, magnetic field, light or by any other crosslinkers.

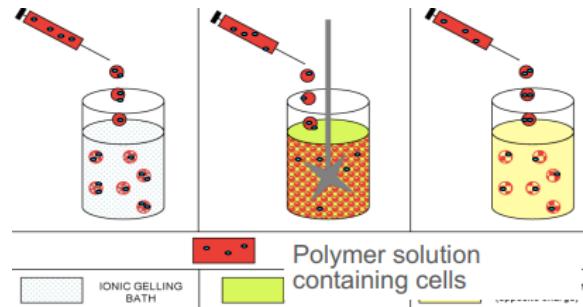


Figure 8.19: (a) dropping the polyelectrolyte solution into a solution of small ions; (b) via a water in oil emulsification technique; and (c) complexation of oppositely charged polyelectrolytes by mixing, with additional coating procedures

8.4 Electro-hydrodynamic jetting (EHDJ)

In the electro-hydrodynamic systems a solution is fed through a positively charged metallic needle. The solution reacts to the presence of the charge, generating repulsive coulombic forces on its surfaces, causing the deformation of the meniscus at the tip of the needle into a Taylor cone. If the voltage is high enough, the electrostatic repulsion on the surface can overcome the surface tension at the apex of the liquid cone, leading to its disintegration (Rayleigh limit) and creating a jet of drops. There are some parameters that we have to assess:

- Speed;
- Starting concentration of polymer;
- Starting concentration of cells;
- Type of polymer.

Taking in consideration these parameters we are able to control the final number outside of cells onto the beads. We can also check the quality of the concentration using a confocal microscope with live/dead staining 8.20. With this process is possible to check how many cells are alive and assess an optimal number of cells.

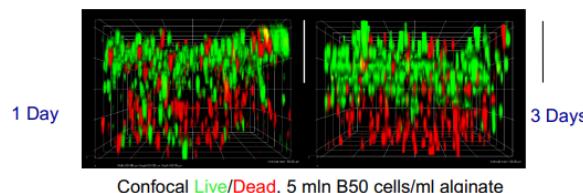


Figure 8.20

8.5. ORGAN PRINTING - BIOPRINTING

8.4.1 Cell encapsulation requirements

- The encapsulation material (polymer) should permit the free passage of nutrients and oxygen (in) and waste products (out) as well as of therapeutic protein products.
- Encapsulation material should prevent high Mw molecules, antibodies and other immunogenic moieties from contacting the encapsulated cells.
- The encapsulation material should protect cells from mechanical stresses and from the host's immune-response.
- The encapsulation material and method should not damage cells neither affect their behavior, as related to the desired function/application.

8.4.2 Cell encapsulation applications

Possible applications for cells encapsulation are drug delivery, bioartificial organs and tissue engineering.

- The encapsulation of therapeutic cells permits the implantation of allogenic and xenogenic cells for the regulation of certain physiological processes damaged by death or senescence of host tissues.
- Microcapsules injected at the transplantation bed allow the release of biomolecules produced by the encapsulated cells, such as insulin produced by encapsulated beta-cells for diabetes type I therapy, pro-angiogenic or anti-angiogenic growth factors to enhance or inhibit vascularization.
- Microencapsulated cells can be used as building blocks for the fabrication of tissues/tissue precursor in vitro or implanted in vivo for tissue regeneration.

8.5 Organ Printing - Bioprinting

8.6 Short-term encapsulation effect

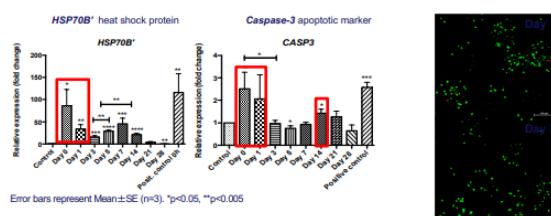


Figure 8.21

- The main function of HSP proteins consist in cells protection against apoptosis, necrosis, hypoxia or any other type of stress.
- Elevated co-expression of HSP70B' and caspase-3 genes points to mild stress condi-
- tions, and cells protection activity.
- The results were supported with Live/Dead test, where we can observed cell death within 48 hours after encapsulation

Chapter 9

Vascularization

9.1 Angiogenesis and pore size

Angiogenesis is the bottleneck of tissue engineering. In order to regenerate functional tissues, stimulation and control or inhibition of angiogenesis are critical. The porosity of the scaffold is fundamental for promoting vascularization. A broad distribution of pore sizes and pore shapes ranging from very small to very large can be required.

In fig 9.1 we have the design of a very simple porous scaffold. Building technique: combine the polymers with spheres, with holes filled with water and then emptied. The ratio between spheres and solution has to be tuned carefully. Black holes are interconnections. Another technique is salt leaching, where we first add crystals to polymeric solution and then salt is removed. The regular shape is kept fixed thanks to crystal. In the case of spherical pores, the interconnection depends on the number of spheres per volume. We can modulate size and interconnection by selecting the concentration.

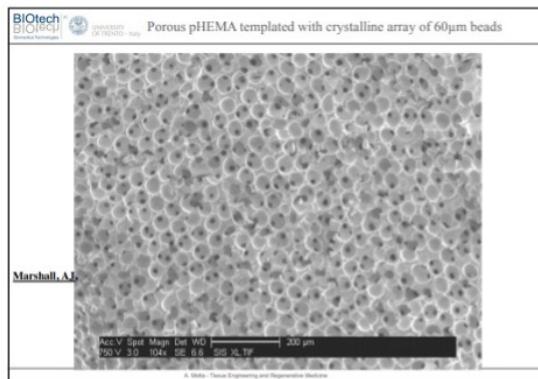


Figure 9.1

If we analyze an image at higher magnification, we observe that black dots measure around 5 micron. By playing with the scaffold, different families can be obtained. Results: more capillaries in the case of smaller pores. Ideal range: 40-50 microns. The highest vascular density was observed

9.2. SCAFFOLD VASCULARIZATION STRATEGIES

in the case of smaller porosity (0-100 micrometers). The range was explored in depth, 20 micron something appeared, the highest is around 40 micron. Very large porosity leads to the formation of a layer of endothelial cell, no 3D tube required for vascularization.

9.2 Scaffold vascularization strategies

1. growth factor delivery: delivery of single or multiple angiogenic GF to stimulate. loading molecules can be performed by mixing with the polymer (fast release), include them in a drug releasing system (polymeric nanoparticle, driven by degradation time of nanoparticles) or link the molecules to the polymeric chain. Which is the rational of the choice? It mainly depends on required timing. Strategies can be also combined in order to release our molecule at different times.
2. cell transplantation: implant EC on the tissue
3. in situ vascularisation with endogenous cells
4. scaffold pre-vascularization: pay attention to the protocol used. We must be sure that robust tubes are produced in order to anastomize with the osteoblasts. We can also include something to attract already existing vascular tubes to the scaffold.
5. decellularised scaffold: decellularised myocardium, not populated by endothelial cells. We take a piece of myocardium, decellularize but leave the basic structure, including tube. The main difficulty in this strategy is to guarantee a very low amount of DNA (check that no debris is left after decellularization), avoid thrombogenesis through forced migration to reform endothelium.
6. angiogenic biomaterials: the best option if possible, design biomaterials able to induce vascularization per se. We can call them “precision biomaterials”, as they are designed according to have a specific function. Nature derived biomaterial: e.g. protein derived from silk, intrinsic angiogenic potential. Synthetic functionalized biomaterials.
7. microfabrication methods: top-down approach, answer to build up vessels and around them the tissue. We can bypass the issue of decellularize by directly inserting endothelial cells.

9.3 Angiogenic potential evaluations

In vitro procedures should be carefully designed, we need to make sure that we are obtaining useful considerations on the scaffold. Fig 9.2 : seed mononuclear cells inside the hydrogel upon isolation of two main endothelial progenitor cells populations from peripheral blood. Impact on the matrix: cells were seeded and checked after 4 and 10 days. In the case of IKVAV modified culture we have more signals. This means we have more cells, more clusters and more layers (migration). The aim was to design a precision material to be used in brain, laminin derived (brain ECM). Strategy: isolate peptide involved in differentiation and link it to polymeric chain. We expect to obtain tubes in angiogenesis, but here we only observe a bunch of cells. What was wrong is the in vitro model, too far from physiology because only one layer is not good. The model was then changed: in order to reach a tube more cells are required e.g. cells able to drive tube formation -> bone marrow stem cells, which differentiate into pericytes, secrete angiogenic factors e.g. VEGF and Ang-1 and produce collagen I and IV for the ECM.

9.4. VASCULARIZATION STRATEGIES

For stem cells isolated from donors we need to at least use 5 replicates, as there is a huge variability. By applying this procedure, we can see a nice formation of tubes (well branched and interconnected) in donor 1, only pieces in donor2 and 3. Takehome message: take into consideration personalized solution in order to avoid variability and to achieve the best result. Secondly the result in co-culture was identical in terms of ECM. If the culturing conditions are good and we introduce SC, we can successfully reach vascularization.

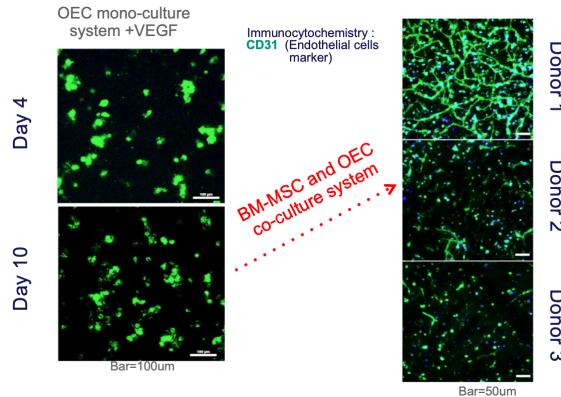


Figure 9.2

Considering the bone, the presence of two families leads to best results: osteoblast work will lead to an increase endothelial cells formation. Osteoblast cells know how much and when to release factors. Osteoblasts also produce a collagenic template used by endothelial cells to form tubes. When we take into account the scaffold, we can support the direct interaction with both osteoblast and endothelial cells. Example: fibroin micronet, co-cultured, 3d scaffold. The endothelial cells become tubes, there is a collagenic network. We need that the tubes are stable. In the microscope view we can see that there is a very physiological structure.

9.4 Vascularization strategies

New strategies to achieve better blood vessels: architecture is important, should provide something to make endothelial cell grow mechanism of angiogenesis can be amplified. Porous matrices can be loaded with growth factor, functionalization to recruit blood vessels from surrounding tissues. They should be able to penetrate and form interconnected tissue.

The main vascularization strategies are:

- angiogenesis: ingrowth of vascular sprouts from the host microvasculature into implanted tissue construct, which finally form a new microvascular network
- inosculation: pre-created vessels in vitro (network, not real vessels) is generated within a tissue construct prior to implantation. Connection with host vessels + already included vessels. Usually animal cells are used, in order to distinguish which vessels are new and which are implanted through staining. Sometimes scaffold tubes are not able to anastomize and die, our aim is to preserve the mixture and anastomize (circle in the picture).

9.4. VASCULARIZATION STRATEGIES

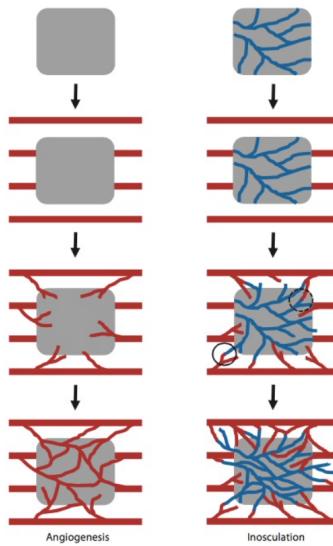


Figure 9.3

9.4.1 Immune response and vascularization

We have also to remember that inflammation (immune response) induces to angiogenesis. Initially we have the recruitment of macrophages. Granulation tissue must be vascularized. Two different structures with different porosity. In the first case (high porosity) we have a fine granulation tissue, angiogenic factors and new vessels. After 2 weeks we have less, much less inflammatory response. In the second case there is no vascularisation, so this is not good. There is a thick layer of fibrotic tissue.

In vivo vascularisation scaffold induced Polymer: form a sponge with a porosity with different pore sizes.

- PDLLA: blood vessels are just around (3wks)
- PDLLAA + silk fibers: very nice capillary branch formation into the scaffold (3wks)

After 6 weeks Silk seems to be intrinsically angiogenic. We need early and efficient angiogenesis, the winner is always the scaffold that achieves the result in a short time. Which are the intrinsic angiogenic properties of silk fibroin net and mechanism? We don't know yet for sure, but we need them!

9.4.1.1 Different polymers in co-culture

Comparison of different polymers with co-culture (HA, calcium phosphate, Ni-Ti, fibroin). The results are quite different, it's important to carefully design the in vitro test, but it is not enough if the substrate is not specifically bioactive.

9.4.1.2 Angiogenesis driven by inflammatory cells

The fibroin net was implanted with and without pre-seeding osteoblasts. A very nice and intense angiogenesis was observed thanks to the interaction between osteoblasts and inflammatory signal.

9.5. INDUCTION OF VASCULARIZATION IN TE SCAFFOLDS

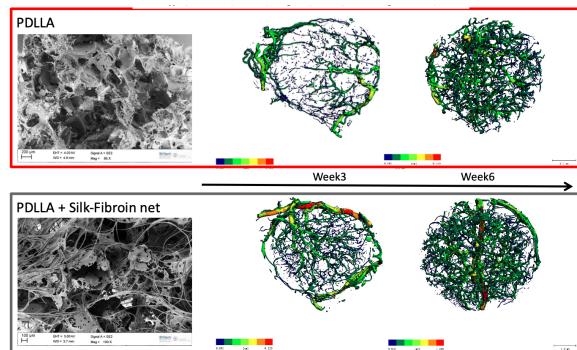


Figure 9.4

Different pre- culture time Manage the inflammatory response better in the case B. Impact on angiogenesis: second scaffold has a better angiogenesis! Drug release system works and speeds up the angiogenesis process.

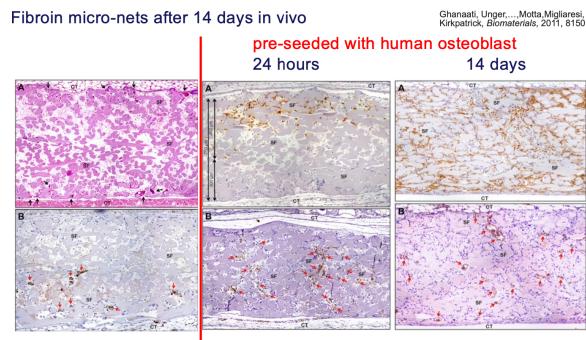


Figure 9.5

9.5 Induction of vascularization in TE scaffolds

After the selection and dosage of the molecules we need to design how to release them into the system.

9.5.1 PDGF-containing microspheres

Dual factor delivery from degradable scaffolds for de novo blood vessels synthesis: we can have PDGF-containing microspheres: the molecules must exit from the vesicle, move in a matrix and then release (very slow process). We need to assess degradation does not occur prior to release and whether microparticles are active and alive throughout the waiting time.

9.5.2 Cytokine delivery

The sequential delivery of VEGF and PDGF-BB using a controlled release polymeric device subcutaneously and in hind limb ischemia model induced a mature new vascular network with vessels

9.5. INDUCTION OF VASCULARIZATION IN TE SCAFFOLDS

having a thick coat of smooth muscle cells. The combination of the two gives the best and fastest angiogenesis.

9.5.3 Microparticles

Scaffold design by microparticle assembly. Microparticles preparation:

- formation of microparticles via single emulsion
- double emulsion: with growth factors

It includes the microspheres: loaded or not with drugs. Sintering: formation of microparticles with some porosity inside Printing: print microparticles Single emulsion: method of sintering.

Very good adhesion with sintering process. Sintering is usually used for ? alloy, but also with polymers and ceramics. The porosity can be improved by mixing with water (water soluble or insoluble materials). You can also use polymers with this technology, at first this technology needed really high temperatures and so not good with polymers, but now lower temperatures are needed.

9.5.4 Strategies for enhancing vascularization

By host: with growth factors pros: easy to engineer, high quality vessels, cons: too slow Really slow is a cons because the tissue needs to be vascularised in order to work properly. Engineered vascular network: pros: immediate perfusion cons: hard to engineer, must be compatible with blood We want to provide porosity but also canals. Also tubes can be provided, for example tubes with ice, to be filled with vessels.

Chapter 10

Bioreactors

10.1 Scaffold development process

We start from a pathology, analyze the microenvironment (not only chemical signals but also mechanical signals, ability to heal) and the tissue to regenerate. From this info we can build scaffold able to mimic the morphology. The goal must be reached by carefully designing materials and manufacturing methods. E.g. 3D printing layer by layer allows us to obtain gradients with different properties. After obtaining a scaffold we need to characterize it with biological function. Once a good result is obtained we can pass to clinical trials: first of all make sure that we are not producing damages to the body, secondly demonstrate suitability for therapeutic purposes.

10.2 3D structures

Growing cells on flat surfaces is unnatural and artificial, it does not make sense in a biological perspective. Natural ECM plays an important role in regulating cellular behaviours by influencing cells with biochemical signals and topographical cues. In 3D cultures, we can control scaffold morphology, architecture and chemistry, bio recognition signaling, degradation mechanisms, patterns; cells behave and respond to stimuli more like they would do *in vivo*.

2D culture substrates are not able to reproduce the complex and dynamic environments of the body, forcing cells to adjust to an artificial flat, rigid surface. 3D matrices or scaffolds are porous substrates that can support cell growth, organisation, and differentiation on or within their structure. Architectural and material diversity have much more impact on 3D matrices than on 2D substrates. Other than physical properties, chemical and biochemical modification with specific biological motives to facilitate cell adhesion, cell mediated proteolytic degradation and growth factor binding and release.

In order to build an effective scaffold we must be very precise and stick to biological processes. While working *in vitro* we must carefully choose cells e.g. monocytes to evaluate immune response. The mechanics should be dynamic, not static. We need to decide stresses to apply, their intensity and timing. Bioreactors should always be designed by keeping in mind the application.