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Innovative strategies for intervertebral disc regenerative medicine: from cell therapies to multiscale delivery systems

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1 **1. Abstract**

2 As our understanding of the physiopathology of intervertebral disc (IVD) degeneration has improved,
3 novel therapeutic strategies have emerged, based on the local injection of cells, bioactive molecules, and
4 nucleic acids. However, with regard to the harsh environment constituted by degenerated IVDs, protecting
5 biologics from in situ degradation while allowing their long-term delivery is a major challenge. Yet, the
6 design of the optimal approach for IVD regeneration is still under debate and only a few papers provide a
7 critical assessment of IVD-specific carriers for local and sustained delivery of biologics. In this review, we
8 highlight the IVD-relevant polymers as well as their design as macro-, micro-, and nano-sized particles to
9 promote endogenous repair. Finally, we illustrate how multiscale systems, combining in situ-forming
10 hydrogels with ready-to-use particles, might drive IVD regenerative medicine strategies towards
11 innovation.

12

13 **2. Outlook section**

- 14
- 15 • Intervertebral disc (IVD) regenerative medicine could benefit from the emergence of cell,
16 molecule-, gene- and siRNA-based strategies.
 - 17 • Design of IVD-specific carriers to protect biologics from in situ degradation, while allowing their
18 local and sustained delivery, is a major challenge.
 - 19 • A critical assessment of IVD-relevant natural and synthetic polymers, and their design as macro-,
20 micro-, and nano-sized particles is provided.
 - 21 • Multiscale systems, combining in situ forming hydrogels with ready-to-use particles, are
22 introduced as new opportunities to stimulate IVD regeneration.

23 **3. Introduction**

24 Low back pain (LBP) is a common health condition and it is now well established that over 80% of adults
25 will experience LBP during their lifetime. LBP has become the second most frequent cause of hospital
26 visits and the leading cause of years lived with disability (Maher et al., 2017; Vos et al., 2016). Thanks to
27 a better understanding of the etiology of LBP, the involvement of intervertebral disc (IVD) degeneration,
28 referred to as disc degenerative disease (DDD) (Vos et al., 2016), was identified.

29 Current treatments of LBP include nonpharmacological, pharmacological, or surgical approaches, all
30 aimed at controlling pain. For early degenerative stages, pharmacotherapies are mainly used (Abdel
31 Shaheed et al., 2016) in association with rehabilitation or education of the patient (Latimer et al., 2004).
32 For more advanced cases, surgical approaches are considered if conservative methods fail. Spine fusion
33 aims to directly remove the IVD and immobilize the adjacent vertebrae. However, this intervention
34 reduces spine mobility and neighboring segments can be affected by an accelerated degenerative process
35 (Baliga et al., 2015). Arthroplasty, with *Nucleus pulposus* or IVD total replacement, aims at conserving
36 the spine motion notably at the affected levels. These interventions, although generally clinically effective
37 in alleviating pain (Galbusera et al., 2008), are very invasive and often include a long postoperative
38 recovery and a non negligible risk of complications (Baliga et al., 2015).

39 Recent progress in our understanding of IVD physiopathology has led to new approaches for IVD
40 regenerative medicine. In view of the decreasing number of cells in degenerating IVD, cell-based
41 therapies were studied with growing interest. However, owing to the restrictive regulatory framework and
42 the difficult clinical translatability of cell-based therapies, the use of biological factors targeting IVD
43 degenerative processes was also contemplated. More recently, the role of gene mutations or microRNA
44 dysregulation was identified at the onset of DDD, leading to the development of a third therapeutic
45 approach based on the delivery of genes or small interfering RNAs (siRNA). In this context, this review
46 will first briefly give recent insights into the pathophysiological mechanisms of DDD and then focus on
47 describing cell-, molecule-, gene-, and siRNA-based strategies for tackling DDD as well as the scaffolding
48 biomaterials used for their respective delivery.

49

50

51 **4. IVD degeneration physiopathology**

52 **4.1. IVD structure**

53 IVDs are fibro-cartilaginous tissues located between the vertebrae that are avascular and non-innervated
54 (Whatley and Wen, 2012). They act as shock absorbers by distributing the mechanical loads along the
55 spine and facilitate trunk mobility (Whatley and Wen, 2012; Zhou et al., 2014). IVDs are generally
56 presented as a tripartite anatomical entity composed of the following parts (Figure 1).

57 **The *Nucleus pulposus* (NP)** is the inner part and the most hydrated region of the IVD (Fontana et al.,
58 2015). NP cell density in human is the lowest of the IVD (Maroudas et al., 1975). These cells can be
59 classified into two types: notochordal cells and nucleopulpyocytes (Colombier et al., 2014). Human adult
60 NP cells produce a highly hydrated extracellular matrix (ECM), rich in proteoglycans (mainly aggrecan)
61 and type II collagen.

62 **The *Annulus fibrosus* (AF)** is the outer part of the IVD. It can be divided into the outer AF containing a
63 oriented lamellar array of densely packed collagen fibers, while the inner AF is less dense and less
64 organized (Raj, 2008; Whatley and Wen, 2012). AF cells have a fibroblastic morphology (Sakai and Grad,
65 2015). The AF is composed of water, types I and II collagen fibers, proteoglycans, and other proteins
66 (Roughley, 2004). The AF retains the pressurized NP and allows for disc compression/decompression
67 (Fontana et al., 2015).

68 **The endplates** (EPs), composed of hyaline cartilage at the top and bottom of the vertebral bodies,
69 interface the IVDs and the adjacent vertebrae. EPs cells are chondrocytes that synthesize an ECM rich in
70 proteoglycans, types I and II collagens, and water. Mechanically, EPs support significant loads during
71 daily-living activities and distribute the intradiscal pressures onto the adjacent vertebrae. They represent
72 the main route for glucose and oxygen to reach the disc cells through the vertebral capillaries and for
73 waste products to be eliminated from the disc (Fontana et al., 2015; Lotz et al., 2013).

74

96 **4.2. Extracellular matrix anabolism/catabolism homeostasis**

97 DDD has a complex and multifactorial etiology. Genetics, environmental causes, mechanical factors,
98 aging, as well as systemic and toxic factors (Elmasry et al., 2015; Hadjipavlou et al., 2008) were identified
99 as risk factors. Interestingly, recent data highlighted that the concomitant presence of notochordal cells
100 and nucleopulpocytes was a pivotal factor for IVD ECM homeostasis (Erwin et al., 2006). Indeed, these
101 two cell types interact via synthesized molecules and the early disappearance of notochordal cells is
102 associated with a downregulation of nucleopulpocyte survival. This decrease in nucleopulpocyte density
103 leads to a misbalance in the ECM anabolism/catabolism ratio that culminates in the alteration of the
104 mechanical properties of IVDs (Erwin et al., 2011). In the advanced stages of IVD degeneration,
105 proteoglycans, hydration, and the type-II/type-I collagen ratio are decreased (Russo et al., 2016; W.-J.
106 Wang et al., 2015), leading to a dehydrated and fibrotic NP with a solid-like mechanical behavior (Russo
107 et al., 2016). Additionally, the AF structure is disorganized and EPs calcify (Fontana et al., 2015; W.-J.
108 Wang et al., 2015). As the degeneration worsens, innervation and vascularization of the disc occur
109 (Whatley and Wen, 2012), leading to discogenic pain (Alkhatib et al., 2014).

110 From a molecular point of view, IVD degeneration is associated with an upregulation of pro-inflammatory
111 cytokines such as interleukin-1 β (IL-1 β) and tumor necrosis factor- α (TNF- α) (Sakai and Grad, 2015).
112 These two molecules trigger the over-expression of matrix metalloproteinases (MMPs) notably MMP-1, -
113 2, -3, -7, -8, and -13. These MMPs are known for their proteolytic action toward collagen and
114 proteoglycans (Kepler et al., 2013b; W.-J. Wang et al., 2015). Similarly to MMPs, there is a marked
115 increase in the a disintegrin and metalloprotease with thrombospondin motifs (ADAMTS) 4 and 5.
116 ADAMTS 4 and 5 act as aggrecanases, thereby contributing to worsening ECM degradation (Molinos et
117 al., 2015; W.-J. Wang et al., 2015). As a pro-inflammatory cytokine, IL-1 β up-regulates the expression of
118 vascular endothelial growth factor (VEGF), nerve growth factor (NGF), and brain-derived neurotrophic
119 factor (BDNF), thereby triggering the neo-vascularization and neo-innervation of IVD that ultimately
120 result in inflammation and pain (Alkhatib et al., 2014; Whatley and Wen, 2012). Finally, in the presence
121 of TNF- α or IL-1 β , NP cells tend to secrete chemo-attractant molecules including (C-C motif) ligand
122 5/regulated on activation of normal T cells expressed and secreted (CCL5/RANTES) or chemokine (C-X-
123 C motif) ligand 6 (CXCL6) (Grad et al., 2016), thus increasing the migration of mesenchymal stem cells
124 (MSCs) (Pattappa et al., 2014). This increased secretion of CCL5 was correlated with painful degenerated
125 IVDs (Kepler et al., 2013a). In addition, stromal cell-derived factor-1 (SDF-1) was over-expressed in
126 degenerated IVDs and involved in NP cell apoptosis (Z. Liu et al., 2016). Interestingly, SDF-1 also
127 demonstrated a chemo-attracting effect toward MSC migration in an ex vivo IVD culture model (Pereira
128 et al., 2014).

129 In contrast to these catabolic factors, anabolic factors stimulate the biological cascade involved in
130 regenerative processes. Among them, Transforming Growth Factor- β (TGF- β) is known to play a pivotal
131 role in collagen and proteoglycan synthesis and in ECM homeostasis (S. Wang et al., 2015). TGF- β is also
132 one of the most potent inhibitors of TNF- α -induced upregulation of MMPs as well as of aggrecan and
133 type-II collagen degradation (Yang et al., 2015). In parallel, Bone Morphogenic Proteins (BMPs) are
134 known to increase proteoglycan and type-II collagen synthesis as well as cell proliferation and
135 differentiation (S. Wang et al., 2015). Among them, Growth and Differentiation Factor-5 (GDF-5 or
136 BMP-14) plays a crucial role during musculoskeletal development and postnatal IVD maintenance (Feng

137 et al., 2015). GDF-5 gene mutations notably lead to disc abnormalities in mice (Li et al., 2004). GDF-5
138 also upregulates the proliferation and secretory activities of NP and AF cells (Chujo et al., 2006) and
139 modulates MMPs, IL-1 β , or TNF- α expression levels (Enochson et al., 2014; Gruber et al., 2014a).
140 Altogether, these data highlight the pivotal role of GDF-5 in IVD matrix homeostasis.

141 The ECM homeostasis of IVDs and the consequences associated with a homeostasis rupture are illustrated
142 in Figure 2.

143

144 **5. From physiopathology to IVD specific biotherapies**

145 Regarding the multifactorial nature of DDD, three main therapeutic strategies were considered in the past
146 few years: cell-, molecular-, and nucleic acid-based therapies (Figure 3).

147 **5.1. Cell-based therapies**

148 A converging body of evidence suggested that IVD degeneration originates from the early disappearance
149 of IVD resident cells. Supplementation of the IVD with regenerative cells thus naturally emerged as a
150 strategy. Repopulating a degenerated IVD with healthy cells aims at either (i) restoring tissue homeostasis
151 by transplanting cells able to secrete a functional ECM or (ii) tackling in situ inflammation by
152 transplanting stem cells with immune-suppressive and anti-inflammatory properties. To achieve these
153 ambitious and clinically relevant objectives, various cells sources were considered. Logically, NP cells
154 harvested from surgically removed IVDs were the first to be used, with promising results in a canine study
155 and in human clinical trials (Meisel et al., 2007). However, their further use was dramatically hampered
156 by the low cell number harvested (Sakai and Grad, 2015). In parallel, pluripotent stem cells such as
157 embryonic stem cells could be of great interest in view of their theoretical potential to differentiate into
158 NP cells. Nevertheless, legal and ethical controversies limit the use of these cells for clinical application.
159 Induced pluripotent stem (iPS) cells were also considered for various applications in regenerative
160 medicine (Hirshi et al., 2014) and their suitability for IVD repair is being studied (Chen et al., 2013) but
161 further studies are needed before their potential can be judged. Last but not least, adipose- (ASCs) or bone
162 marrow-derived (MSCs) mesenchymal stem cells in their undifferentiated state are likely the most widely
163 used in the DDD context. In addition to providing us with promising results (see next section), the
164 extensive use of MSCs raised the question of their allogeneic or autologous origin (Richardson et al.,
165 2016). Indeed, MSCs are considered hypo-immunogenic and immune-evasive cells able to be transplanted
166 across major histocompatibility barriers (Ankrum et al., 2014), making the use of allogeneic MSCs a
167 credible alternative to autologous ones.

168 ASCs are considered particularly advantageous because of their abundance in fat tissue, their accessibility,
169 and their more potent immuno-modulatory effects in comparison with MSC isolated from bone marrow
170 (Li et al., 2015). Of interest, human ASCs exhibit a nucleopulpogenic potential when committed with
171 TGF- β 1 and GDF-5 (Colombier et al., 2016). When implanted in mice, these cells also demonstrated their
172 ability to secrete GAGs and type II collagen, consistently with native NP cell capacities (Colombier et al.,
173 2016). In rabbits, undifferentiated MSCs injected in the NP inhibited disc degeneration by reducing MMP-
174 2, MMP-3, and MMP-13 expression and by promoting type-II collagen synthesis, resulting in the
175 maintenance of the IVD structure (Miyamoto et al., 2010). The beneficial outcomes of injecting
176 undifferentiated MSCs were further demonstrated in other animal models, such as rat, mouse, dog, or pig
177 models (Henriksson et al., 2009; Hiyama et al., 2008; Jeong et al., 2009; Marfia et al., 2014; Richardson et
178 al., 2016). In these studies, IVD degeneration was significantly slowed down, disc height and MRI signal
179 intensity improved, and nucleopulpogenic ECM neo-synthesis was observed. In humans, the intradiscal
180 injection of MSCs could lower pain and disability (Orozco et al., 2011). This relief was associated with
181 the anti-inflammatory (Maria et al., 2016; Sakai and Grad, 2015) and pro-anabolic properties of MSCs
182 (Shabbir et al., 2009). Several ongoing clinical trials are conducted worldwide to determine the optimal
183 cell source (autologous or allogeneic) to be used in DDD (Sakai and Andersson, 2015).

184 Despite their global promise, the intradiscal injection of “naked” MSCs remains hampered by the harsh
185 environmental condition the cells experience when injected into a degenerated IVD (low oxygen tension,
186 low nutrient level, metabolic product accumulation, acidic pH, high osmolarity, and high intradiscal
187 pressure). These stringent conditions are responsible for the poor survival rate and altered activity of the
188 injected cells. To overcome this hurdle, tissue engineers proposed to use cytoprotective 3D scaffolds able
189 to create a biomimetic environment that could support cell survival. These scaffolds could additionally
190 promote cell retention at the injection site thereby limiting the risk of off-target effects related to cell
191 effusion. These scaffolds will be detailed in the last section of this review.

192

193 **5.2. Molecular therapies**

194 More recently, and facing the numerous limitations of cell transplantation, an alternative to cell-based
195 therapies emerged. It consists in exploiting the ability of chemoattractant molecules to mobilize and
196 recruit endogenous progenitor cells at the degenerated site (Illien-Jünger et al., 2012). This strategy, based
197 on the capability of the endogenous body to self-regenerate, could free clinicians from the restrictive
198 regulatory framework and cost ineffectiveness of cell transplantation. With respect to the myriad of factors
199 playing a crucial role in IVD physiopathology, many chemoattractant, pro-anabolic, and anti-catabolic
200 molecules were contemplated for the stimulation of resident cells or the attraction/recruitment of
201 endogenous progenitors.

202

203 One key factor for improving the endogenous repair of IVDs via cell attraction/recruitment is the presence
204 of endogenous progenitors with stem cell properties in the IVD (Risbud et al., 2007). These IVD-derived
205 progenitor cells (either from NP or EPs) are very similar to MSCs isolated from the bone marrow of the
206 same patient (Blanco et al., 2010). Interestingly, analyses of the migration capacity of progenitors located
207 in the NP, AF, and EP indicate that EP progenitors have the highest migration and invasion potency (S.
208 Liu et al., 2016). To favor cell migration in degenerative IVDs, chemokines were studied since, as
209 chemotactic cytokines, they are involved in cell mobilization, attraction, and homing. Regarding IVDs, the
210 three main chemokines expressed are SDF-1, CCL5, and CXCL16, both in healthy or degenerated IVDs.
211 Interestingly, the expression of SDF-1 and CXCR4 was upregulated in degenerated IVDs, notably in
212 MSCs isolated from the EPs (S. Liu et al., 2016; Zhang et al., 2014). SDF-1 also increased MSC attraction
213 in a nucleotomized disc model (Pereira et al., 2014). The implication of SDF-1 in stem cell attraction was
214 also demonstrated in many tissues including myocardium or periodontal ligament (Kaku et al., 2017;
215 Purcell et al., 2012). Recently, IVD cells stimulated with IL-1 β and TNF- α were found to secrete a higher
216 concentration of CCL5 compared with IVD cells cultured under physiological conditions (Gruber et al.,
217 2014b). The presence of CCL5 in degenerated IVDs as well as herniated and scoliotic discs was further
218 demonstrated by proteomic assays in other in vitro studies in humans (Kawaguchi et al., 2002; Kepler
219 et al., 2013a). Finally, in AF cells, CXCL16 is up-regulated by IL-1 β , suggesting a role for the
220 CXCL16/CXCR6 system (Gruber et al., 2016).

221 Altogether, these data highlight the role of chemokines in the degenerative process of IVDs and strengthen
222 their relevance in developing new strategies to attract progenitor cells and stimulate endogenous repair
223 (Illien-Jünger et al., 2012).

224

225 In line with our understanding of DDD, anabolic factors can contribute to IVD regeneration. A large panel
226 of anabolic factors was contemplated for IVD regeneration. Here, we mainly focus on the factors that
227 demonstrated promising results in several reports.

228 TGF- β contributes toward maintaining the expression levels of connective tissue growth factor (CCN2),
229 one of the most potent factors able to stimulate aggrecan synthesis, in NP cells (Tran et al., 2014). In
230 addition, CCN2 suppressed the inductive effect of IL-1 β on catabolic genes (Tran et al., 2014).

231 Considering that pro-inflammatory cytokines decreased CCN2 expression, this makes TGF- β an anabolic
232 factor able to counteract the effects of pro-inflammatory cytokines (Bedore et al., 2014). TGF- β was also
233 used in association with dexamethasone to stimulate the proliferation of degenerated human NP cells and
234 exhibit an anti-catabolic gene expression profile (Liang et al., 2013).

235 GDF-5 is a major factor involved in early IVD development, and mutations of its related gene lead to
236 severe structural impairment (Feng et al., 2015). Noteworthy, GDF-5 is down-regulated by IL-1 β and
237 TNF- α . In vitro, GDF-5 favors cell proliferation and stimulates the synthesis of proteoglycan and type II
238 collagen, thus demonstrating its anabolic activity (Chujo et al., 2006). GDF-5 was used to differentiate
239 human ASCs into nucleopulpcytes, in association with TGF- β 1 (Colombier et al., 2016). Interestingly,
240 intradiscal injection of GDF-5 in rats and rabbits was shown to increase disc height. Consequently, clinical
241 trials were conducted to evaluate the safety and efficacy of intradiscal injection of GDF-5 in patients with
242 DDD (NCT01158924, NCT00813813, NCT01182337, and NCT01124006 on clinicaltrials.gov). As far as
243 we know, no peer-reviewed publication is currently available. While a modest improvement in clinical
244 parameters was observed, some adverse events were also reported suggesting that despite the relevance of
245 GDF-5 in DDD, its direct intradiscal injection may not be the most suitable route.

246 Other anabolic factors were investigated and their anti-apoptotic activity toward NP cells was evidenced.
247 Indeed, aside from promoting angiogenesis, it is now clear that in IVD, vascular endothelial growth factor
248 (VEGF) also prevents NP cell apoptosis (Tran et al., 2010). Additionally, insulin-like growth factor (IGF-
249 1), platelet-derived growth factors (PDGFs), epidermal growth factor (EGF), and fibroblast growth factor
250 (FGF-2) stimulate cell proliferation and inhibit cell apoptosis in human and animal cells (Vasiliadis and
251 Pneumaticos, 2014). Together with the aforementioned CCN2 (Tran et al., 2014), these factors
252 demonstrated their anabolic activity.

253 Regarding the role of anabolic factors in DDD treatment, the use of platelet-rich plasma (PRP), which
254 contains a myriad of multifunctional growth factors, was recently proposed. Two clinical trials suggested
255 the efficacy of PRP in DDD (Kristin et al., 2017; Tuakli-Wosornu et al., 2016). Further double-blind and
256 controlled clinical trials are now required to confirm these first results.

257

258 Recent studies highlighted the key role of both TNF- α and interleukins in inflammation and further IVD
259 degeneration, notably via the up regulation of MMPs and aggrecanases.

260 Thus, anti-TNF- α treatments (infliximab) were proposed and showed a significant decreased expression of
261 IL-1 β and IL-6 in NP cells (Walter et al., 2015), thereby demonstrating their anti-catabolic activity. In
262 addition, the local administration of anti-IL-6 receptor, such as atlizumab or tocilizumab, lowered
263 inflammation-related pain in humans (Sainoh et al., 2016). Interestingly, human IVD cells overexpressing
264 IL-1 receptor antagonist were able to lower an IL-1-dependent increase of MMPs when injected into IVD
265 degenerative explants, demonstrating their anti-catabolic potential (Le Maitre et al., 2007). In parallel,
266 because NF- κ B increases IVD degeneration via the upregulation of pro-inflammatory cytokines (Fontana
267 et al., 2015), studies examined its inhibition. Indeed, human NP cell exposure to NF- κ B inhibitor led to
268 significantly decreased IL-1 β -dependent MMPs and ADAMTS upregulation (Orita et al., 2013; Zhongyi
269 et al., 2015). In addition, inhibition of other cytokines such as BDNF, NGF, VEGF, or TNF- α was studied
270 and showed promising results in the context of inflammation-associated DDD (Orita et al., 2011; Walter et
271 al., 2015; Wu et al., 2011)

272 In parallel to the previously described IVD degenerative processes, aging-associated accumulation of
273 oxidative damage is of great interest. Collagen is subject to glycoxidation reactions that form advanced-
274 glycation end-products (AGEs). The accumulation of these AGEs molecules contributes to IVD
275 degeneration by increasing TNF- α , MMPs, and aggrecanases expression, and to a greater extent AGEs
276 participate in the alteration of structural and mechanical properties of the ECM (Colombini et al., 2008).
277 Interestingly, administration of an anti-inflammatory drug (pentosan-polysulfate) associated with anti-

278 AGE molecules (pyridoxamine) alleviated NP degeneration by lowering MMPs and ADAMTS production
279 in diabetic mice (Illien-Junger et al., 2013).

280

281 With the discovery of molecules involved in IVD degenerative processes, new therapeutic approaches
282 were proposed, either through stimulating anabolism or inhibiting catabolism. Despite encouraging
283 results, the short half-life of such molecules does not allow for long-term activity, and they may not be
284 sufficient to reactivate the physiological IVD machinery. Whether embedding these molecules in
285 biomaterials to allow their in situ sustained and controlled delivery, while protecting them from
286 proteolytic degradation, is a good option will be discussed in the next section.

287

288 **5.3. Gene and interfering RNA therapies**

289 An alternative to biotherapies may be the genetic modification of resident cells through gene transfer, for
290 which two strategies can be envisioned: (i) The vector containing the gene can be introduced in the body
291 (direct or in vivo gene therapy) or (ii) target cells can be harvested, genetically modified in vitro, and re-
292 implanted (indirect or ex vivo gene therapy) (Goldspiel et al., 1993). Considering the decreased number of
293 cells in degenerative IVD, direct gene therapy could be of interest for early degeneration, while the
294 indirect approach seems more suitable for more severe cases. Two carriers were used for gene delivery:
295 non-viral and viral vectors. Non-viral gene therapy refers to the use of naked DNA or cationic compounds
296 (liposomes, polyplexes, lipopolyplexes) (Daraee et al., 2014; Rezaee et al., 2016). Such gene carriers help
297 avoid the risks of mutagenicity, viral infections, and viral immunogenicity (Mintzer and Simanek, 2009).
298 However, their application in IVD degeneration treatment remains limited. Regarding viral vectors, while
299 retroviruses are the most commonly used, they have a more potent action on cells with a high dividing rate
300 making them poorly adapted for NP-cell-mediated IVD gene therapy (Miller et al., 1990). Many authors
301 thus chose adenoviruses despite their potential infectious and immunogenic risk owing to their higher
302 transduction efficiency notably in quiescent or slowly dividing cells. Therapeutic genes encoding for anti-
303 inflammatory agents or MMPs inhibitors (e.g., TIMP-1) were considered as they could drastically reduce
304 the expression of MMPs involved in ECM degradation (W.-J. Wang et al., 2015). It was found that the use
305 of adenoviral vectors coding for BMP-2 and TIMP-1 on human degenerated disc cells increased
306 proteoglycan synthesis (Wallach et al., 2003). Adenoviruses were also used to deliver genes coding for
307 growth factors such as GDF-5 (X.-W. Luo et al., 2016) or TGF- β (Nishida et al., 1999) and they were
308 shown to efficiently stimulated anabolism.

309 In parallel to classic gene therapies, there is growing interest in the use of siRNAs. SiRNAs are short non-
310 coding sequences of 20–25 nucleotides that control gene expression by binding to their target mRNA and
311 causing the repression of RNA translation or degradation (Bartel, 2004). Many miRNAs deregulations
312 emerged as key players in IVD degeneration by triggering NP cell apoptosis, ECM degradation, or NP cell
313 proliferation and inflammatory response (see (Zhou et al., 2017) for review) and can thus be envisioned as
314 new therapeutic targets.

315 Gene- and RNA interfering-based therapies seem to be particularly interesting for restoring the
316 anabolic/catabolic balance in IVDs. However, as NP cell density drastically decreases during
317 degeneration, the effectiveness of such approaches needs to be clarified. Finally, adenoviruses or siRNAs
318 are not integrative tools and do not lead to extended gene expression. In addition, as siRNAs need to be
319 protected from endonuclease degradation, the use of specifically designed biomaterial-based systems, as
320 mentioned previously for peptidic molecules, could be a reasonable option.

321

322 As stated above, the broadening of our understanding of IVD degenerative processes led to new
323 therapeutic strategies based on the local administration of cells, molecules, genes, or siRNA and several
324 clinical trials assessing these three main therapeutic strategies have been reported (Table 1). As of

325 November 2017, eight clinical trials have been published using autologous or allogenic cells, four trials
326 have reported the injection of anabolic factor GDF-5 and only one trial is based on PRP. To the best of our
327 knowledge, there is no ongoing clinical trial based on gene or RNA interfering therapies for IVD
328 regeneration. Despite encouraging results, these therapies need further improvement with regard to the
329 harsh environment in which these biologics are injected. Thus, biomaterials were considered so as to
330 improve cell survival and protect molecules and nucleic acids from degradation. These biomaterials could
331 additionally allow the biologics to be delivered in a spatiotemporally controlled manner thereby promoting
332 long-term IVD regeneration.

333

334 **6. Engineering biomaterials for IVD delivery**

335 Successful delivery of fragile biologics into IVDs is highly challenging, mostly because of their short in
336 vivo half-life. With the aim of targeting this poorly vascularized tissue, delivery routes are limited and
337 thus minimally invasive direct injection received attention. While there are reviews extensively describing
338 pharmaceutical innovations, including self-setting hydrogels, microspheres, and liposomes, for the oral
339 (Diab et al., 2017), the intra-vitreous (Bochet and Fattal, 2012), and the transdermal (Naves et al., 2017)
340 routes, only few papers are available providing a critical assessment of IVD-specific approaches for local
341 and sustained delivery. Here, we highlight the rational selection of natural or synthetic polymers, we
342 exemplify the design of delivery systems based on macro-, micro-, and nano- sized properties, and we
343 illustrate how multiscale combined systems might help direct IVD regenerative medicine strategies.

344

345 **6.1. Natural and synthetic polymers**

346 One key parameter in the design of delivery systems is the selection of the constitutive polymer, since the
347 chemical functionalities of a polymer affect all aspects of its performance, including encapsulation
348 efficiency as well as polymer degradation rate and release (Mitragotri et al., 2014). As such, recent
349 advances in delivery systems are based on polymers tailored for a specific tissue and engineered to exert
350 distinct biological functions.

351 **Among natural polymers**, collagen, hyaluronic acid (HA), alginate, and chitosan were extensively
352 studied (Sionkowska, 2011). Extracted from natural sources or obtained by bacterial fermentation, these
353 macromolecules offer various advantages such as large-scale production, low toxicity, and similarity to
354 ECM components.

355 Collagen, a major component of healthy IVD ECM, exhibits biodegradability, weak antigenicity, and
356 superior biocompatibility compared with other natural proteins. The widespread biomedical use of
357 collagen is based on its ability to be shaped into films, sponges, and micro- or nanoparticles (Liu et al.,
358 2013). Gelatin, derived from collagen hydrolysis, is biodegradable in a physiological environment, with
359 the formation of harmless metabolic products. In degenerated IVD, injectable gelatin microspheres
360 impregnated with PRP reduced NP cells apoptosis and slowed down degeneration in adult rabbits
361 (Sawamura et al., 2009). HA, another leading constituent of the NP ECM, was also studied as a scaffold
362 for tissue engineering (Sionkowska et al., 2016), but less so for IVD strategies.

363 Although they cannot be found in the human body, alginate, a cell wall polysaccharide constituent of
364 brown algae, and chitosan, a polysaccharide synthesized by crustaceans, were used for multiple
365 biomedical applications (Bernkop-Schnürch and Dünhaupt, 2012). Alginate microparticles were
366 designed for the in vitro delivery of PRP in the treatment of degenerated IVDs (Sawamura et al., 2009);
367 however, their ionic gelation-dependent low stability in the body might constitute a drawback. Similarly,
368 cationic chitosan was considered for IVD regenerative applications, as a cell or molecule carrier (Cheng et
369 al., 2010; Roughley et al., 2006).

370 **Synthetic polymers** are obtained through reproducible industrial processes, with tunable physicochemical
371 and mechanical properties. The most commonly used polymers for drug delivery systems are poly(lactic-
372 co-glycolic acid) (PLGA) and poly- ϵ -caprolactone (PCL) (Kondiah et al., 2016).

373 PLGA microspheres were widely used for the release of TGF- β 3 or anti-inflammatory molecules
374 (Bhardwaj et al., 2007; Liang et al., 2013). They were used for recombinant human GDF-5 delivery in
375 vivo in a rat model and increased the synthesis of aggrecan and type II collagen (Yan et al., 2013). PLGA
376 microspheres were also used for the sustained delivery of IL-1 β receptor antagonist in a 3D culture model,
377 and this treatment effectively attenuated the IL-1 β -mediated inflammatory response of the 3D cultured NP
378 cells (Gorth et al., 2012).

379 PCL, a semi-crystalline and hydrophobic polymer, is currently used in FDA-approved sutures for surgery
380 and in commercially available contraceptive devices (Wei et al., 2009). PCL is often associated with
381 hydrophilic polymers, either natural or synthetic ones (e.g., PEG, PAA or pNIPAAm), to form di- or
382 triblock amphiphilic copolymers (Fu et al., 2014; Li and Tan, 2015). Moreover, PCL was extensively used
383 for IVD tissue engineering, especially for AF repair (Wismer et al., 2014). PCL-based or PCL-coated
384 microparticles were successfully designed for the sustained delivery of bioactive molecules, although not
385 for IVD applications (Jung et al., 2015; Z. Luo et al., 2016).

386 In the context of IVD regenerative medicine, the usefulness of these natural and synthetic polymers is
387 broadened further by our ability to create systems with multiple shapes and sizes that deliver drugs,
388 cytokines, growth factors, or stem cells in a controlled manner. They are reviewed in the following
389 sections.

390

391 **6.2. *In situ* forming hydrogels as macro-scaled delivery systems**

392 Hydrogels, networks of hydrophilic polymers capable of absorbing large amounts of water, are of
393 particular interest for highly hydrated NP and AF tissue regeneration. Besides water, they provide
394 mechanical support and a 3D microenvironment adapted for cell survival and proliferation. Attention was
395 focused on the development of stimuli-responsive polymer solutions that could be injected in a minimally
396 invasive manner prior to *in situ* solidifying, ensuring a perfect fit to the desired tissue. The *in situ*
397 formation of hydrogels occurs through chemical covalent bonds or physical bonds (electrostatic or
398 hydrophobic interactions).

399 **Among chemical crosslinking**, photo-induced crosslinking requires the production of radical initiating
400 species and a precursor solution of a photoreactive polymer, usually a polyethylene glycol (PEG) acrylate
401 or methacrylate derivative (Francisco et al., 2014; Li et al., 2014). While the use of photo-crosslinked
402 polymers allows for time-controlled gelation, the limited access of the light apparatus to the target tissue
403 weakens this strategy for IVD regeneration. Click chemistry, Schiff's base chemistry, Michael-type
404 reactions, and enzymatic, protein, or pH modification-based crosslinking approaches might be better
405 suited for IVD *in situ* gelation (Blanquer et al., 2014). These processes are generally compatible with cell
406 or bioactive molecule loading as they often take place in mild conditions (37°C, physiological pH, short
407 period). In the early 2000s, a silanized cellulose-based hydrogel, in which crosslinking is triggered by pH,
408 was developed (Bourges et al., 2002). The importance of this hydrogel has been demonstrated for the
409 regeneration of articular cartilage (Vinatier et al., 2009) and myocardium (Mathieu et al., 2012) and it is
410 currently studied for IVD regeneration (Henry et al., 2017).

411 **Physically crosslinked** hydrogels are formed via non-covalent interactions in response to an external
412 stimulus (pH, temperature, molecule level, light, pressure, electric current) (Qiu and Park, 2012). Thermo-
413 responsive polymers spontaneously undergo a sol-gel transition when exposed to physiological
414 temperature after *in vivo* injection. Recombinant polypeptides [e.g., elastin-like polypeptide (Nettles et al.,
415 2010)], naturally derived polysaccharides [e.g., chitosan (Cheng et al., 2013)], and synthetic polymers
416 [e.g., block copolymers of PEG and PLGA (Tran et al., 2012), poly (N-isopropylacrylamide) (p-

417 NIPAAAM) (Willems et al., 2015)] are the most commonly reported thermo-responsive polymers for IVD
418 regeneration.

419 **In situ-forming hydrogels** exhibit many advantages for IVD treatment. They provide an aqueous
420 environment that can rehydrate the degenerated microenvironment, they protect biologics, and they can
421 restore, at least partially, the mechanical properties of IVDs. The permeability of hydrogels facilitates
422 good transport of nutrients to encapsulated cells. Nevertheless, hydrogels can present some restrictions
423 due to, for example, their weak mechanical properties. One of the main drawbacks of in situ-forming bulk
424 hydrogels is the difficulty to tune their release characteristics. Indeed, while the diffusion of entrapped
425 molecules can be slightly modified by adjusting the polymer network mesh, exchanges with the targeted
426 microenvironment are limited to the interface hydrogel/tissue, usually in the order of a few cm²/g. By
427 comparison, micro- and nano-scale delivery systems can develop areas greater than 1,000 m²/g as
428 measured by N₂ adsorption–desorption experiments.

429

430 **6.3. Micro-scale delivery systems**

431 In parallel to stimuli-responsive polymer solutions ensuring a perfect fit to the IVD defect following in
432 situ gelation, injectable micro-sized particles, with diameters varying from 20 μm to 200 μm, have release
433 kinetics depending on their size and shape. Their release properties can be modified by tuning parameters
434 such as the size, chemical composition, crosslinking degree, or molecular weight of the polymer (Balmert
435 and Little, 2012). Moreover, chemical modification of the microparticle surface can allow for covalent
436 coupling of factors such as antibodies and receptors. Microparticles are able to deliver biological factors
437 over a long period while protecting them from inactivation (Wenk et al., 2009).

438 Many inorganic and polymer-based microparticles can be prepared by simple or double emulsion
439 methods, in which a solution of the material is emulsified in water, followed by the formation of particles
440 during the evaporation of the organic solvent. A recent modified protocol combines a crosslinking process
441 and the emulsion step in an oil phase (Phromsopha and Baimark, 2014). Other production methods are
442 spray drying, gelation, coacervation, electrospray, supercritical fluid mixing, and microfluidic technique.
443 While their main advantage over classic emulsion is the narrower particle size distribution, the yield is
444 often disappointing.

445 Recently, GDF-5-loaded PLGA microspheres, with an in vitro sustained delivery for 42 days, were
446 injected into degenerated rat IVDs. Restoration of the disc height to almost 90% of the non-punctured disc
447 was achieved (Yan et al., 2013). Moreover, the injection significantly increased aggrecan and type II
448 collagen mRNA expression, whereas type I collagen mRNA expression was significantly decreased,
449 suggesting that a micro-sized carrier could be of interest for IVD regeneration (Yan et al., 2013).

450 **6.4. Nano-scale delivery systems**

451 Alongside microparticles, many materials were used to design nanoparticles for biomedical applications,
452 as nanocarriers are easily taken up by cells (Wilczewska et al., 2012). Polymeric nanoparticles, liposomes,
453 dendrimers, polymers, silicon or carbon materials, and magnetic nanoparticles are examples of nano-
454 carriers used as drug delivery systems notably for cancer (Sun et al., 2014) or bone regeneration (van Rijt
455 and Habibovic, 2017). In this section, we highlight the particles that are feasible for use in the context of
456 DDD.

457 **Polymeric nanoparticles:** Bioactive molecules can be either encapsulated or immobilized onto the
458 surface of the nanoparticles, then released by desorption, diffusion, or nanoparticle erosion. Thus,
459 degradable nanocarriers are widely used in biomedical applications as they can be hydrolyzed in situ
460 (Kohane et al., 2006) and often considered to be biocompatible with tissues and cells, since they do not
461 induce pro-inflammatory or immune reactions (des Rieux et al., 2006). An upregulation of aggrecan and
462 type II collagen was achieved by using chitosan/Poly-(c-glutamic acid) nanoparticles loaded with the anti-

463 inflammatory drug diclofenac. These nanoparticles were injected into a needle-punctured bovine IVD
464 cultured in vitro with IL-1 β . Additionally, down-regulation of IL-6, IL-8, MMP1, and MMP3 was noted
465 (Teixeira et al., 2016).

466 **Self-organized spherical liposomes**, usually 80–300 nm in diameter, contain at least one lipid bilayer
467 surrounding an aqueous core (Daraee et al., 2014). They are thus able to encapsulate both hydrophobic
468 and hydrophilic molecules within the lipid bilayer or the aqueous core, respectively. Liposomes were first
469 used to improve the solubility and therapeutic index of chemotherapeutic agents, by allowing targeted
470 delivery and enhanced tissue penetration. Since liposomes interact with cells by adsorption, fusion,
471 endocytosis, or lipid transfer, they were studied as gene and siRNA carriers, especially cationic liposomes
472 (Majzoub et al., 2016). The release kinetics can be tuned by the liposome composition, pH, osmotic
473 gradient, incorporation of temperature- or pH-sensitive lipids (dos Santos Giuberti et al., 2011), and the
474 surrounding environment. Liposomes were thus used for drug delivery in various applications such as in
475 anti-cancer or anti-inflammation therapies or to reach specific organs (lungs, liver, spleen, brain).
476 However, to date, their use for tackling IVD degeneration has not been reported. Interestingly, lipid
477 nanocapsules were successfully designed for nucleic acid delivery (Huynh et al., 2009) and their
478 application in siRNA delivery for IVD regeneration could be of interest.

479 **Dendrimers** are branched polymers whose toxicity is mainly related to functional groups presented at the
480 surface layer and can be lowered by functionalization, e.g., by PEGylation (Zeng et al., 2016). This
481 functionalization improves the surface activity of dendrimers as well as their biological and
482 physicochemical properties. Drugs are entrapped in the internal structure of the dendrimer, adsorbed or
483 chemically bonded onto the dendrimer surface according to its properties (sensitivity to temperature or
484 pH, toxicity, solubility). One of the most commonly used in biomedical applications is poly-(amino
485 amide; PAMAM–) as it was proven to be non-toxic and safe (Chauhan et al., 2009) and its capacity for
486 gene or drug delivery was extensively explored (Zhao et al., 2011). Indeed, PAMAM dendrimers can
487 modulate cytokine and chemokine release (Seelbach et al., 2015) and exhibit anti-inflammatory behavior
488 by inhibiting proinflammatory cytokine secretion (Bosch, 2011). Thereby, dendrimers demonstrated their
489 suitability in rheumatoid arthritis and musculoskeletal diseases and their application in IVD regeneration
490 could be envisioned.

491 **Carbon nanotubes** are the most commonly used carbon nanoparticles and were widely studied for drug
492 delivery in biomedical applications (Beg et al., 2011). They are characterized by a single or multiwall
493 graphene-based tube with a high specific surface area and possibly excellent electronic and thermal
494 conductivity (Beg et al., 2011). They have exceptional mechanical properties and are thus of interest as
495 fillers in composite materials (Harrison and Atala, 2007). Toxicity due to their dimensions and released
496 impurities was reported, but their biocompatibility can be enhanced with chemical surface engraftment of
497 PEG, PAMAM, or albumin (Harrison and Atala, 2007). Molecules are encapsulated inside carbon
498 nanotubes or chemically adsorbed onto their surface. Carbon nanotubes improve the mechanical properties
499 of hydrogels (Dong et al., 2013) and can be of potential interest for IVD regenerative medicine.

500 **Mesoporous silica nanoparticles** are a widely studied biomaterial for their drug delivery capacity in
501 particular for cancer therapies (Yang and Yu, 2016) or nerve growth (Sun et al., 2016). Indeed,
502 mesoporous colloidal silica exhibit several advantages such as cytocompatibility, a high specific surface
503 area, and ease of functionalization (Moritz and Geszke-Moritz, 2015). Additionally, the size, shape,
504 surface pattern, porosity, and pore size of silica nanoparticles are easily tunable (Hoffmann et al., 2006).
505 In the biomedical field, the best known mesoporous silica nanoparticles are MCM-41 and SBA-15. Drug
506 loading into mesoporous silica might occur via chemical or physical adsorption, and release is usually
507 controlled by diffusion throughout the mesopores. The potential of these particles was demonstrated for
508 biomedical applications (Vallet-Regi and Balas, 2008) and more recently for the IVD (Henry et al., 2017).

509 **6.5. Multiscale systems for complex regenerative strategies**

510 Over the past few years, composite materials, composed of at least two distinct components, drew
511 attention in biomedical engineering since they possess additional properties that each constituent alone
512 does not exhibit. They were widely studied for bone regeneration, and successful clinical trials were
513 recently reviewed (Jahan and Tabrizian, 2016).

514 Regarding specific IVD applications, we would like to direct the reader's attention to the multiscale
515 systems illustrated in Figure 4 that emphasize the integration of technologies such as in situ-forming
516 hydrogels with ready-to-use micro- or nanoparticles. The assembly of specific functionalities obtained by
517 combining a bulk matrix, to restore hydration and mechanical properties, with a long-term reservoir of
518 cells or molecules, to provide biological cues, could constitute a breakthrough in IVD regenerative
519 therapies. The main outcome of this review is to highlight the crucial role of biomaterials in IVD
520 regenerative medicine strategies. With regard to the aforementioned device needs, we would like to stress
521 that biomaterials are considered the driving force of the IVD regenerative medicine train (Figure 5).
522 Biomaterials are essential for the design and development of hydrogels capable of restoring load-bearing
523 function while supporting cells until an appropriate ECM is synthesized. Biomaterials are also essential
524 for engineering efficient and robust delivery devices that can be tailored to specific tissues.

525 For example, we recently reported on a multiscale system where an Si-HPMC hydrogel supplemented
526 with rod-shaped silica nanofibers provided sustained release of TGF- β 1 and GDF-5, compared with the
527 plain hydrogel, while maintaining their in vitro biological activity (Henry et al., 2017). Recently, TGF- β 3-
528 loaded PLGA nanoparticles dispersed in a dextran/gelatin hydrogel, although not injectable, increased the
529 gene expression of both aggrecan and type II collagen of MSCs in vitro (Gan et al., 2016). We believe that
530 promising engineering technologies such as interpenetrating network-strengthened hydrogels (Gan et al.,
531 2017), self-assembling nanofibrous peptide hydrogels (Wan et al., 2016), and programmed
532 sequential bioactive factor delivery to target multiple repair steps (Bayer et al., 2017) could lead to
533 significant advances in IVD regenerative strategies. Finally, an ambitious strategy would aim at delivering
534 chemo-attractants from micro- or nanoparticles to attract progenitor cells toward a hydrogel-based
535 microenvironment that would promote endogenous repair.

536

537

538

539 **7. Conclusion**

540 IVD degeneration is a major health concern and today, thanks to recent discoveries, the physiopathology
541 of this disease is better understood. Building on this deeper knowledge, various biotherapies were
542 proposed in the past few years, including cell-based therapies, molecular therapies, and nucleic acid-based
543 therapies. These therapies aim at locally injecting biologics including cells, proteins, or DNA as
544 mentioned earlier. Even if promising results have been obtained in vitro, the harsh environment of
545 degenerated IVDs is a major concern that has hampered the clinical translatability of such concepts. The
546 main outcome of this review is to highlight the essential role of biomaterials in strategies for IVD
547 regenerative medicine. Indeed, injectable in situ crosslinking hydrogels could restore IVD height and
548 hydration and would be particularly suited for IVD regenerative strategies. Additionally, hydrogels can
549 improve cell viability and protect bioactive molecules from in situ degradation. Finally, hydrogels can be
550 used as scaffolds, supplemented or not with micro- or nanoparticles. These so-formed multiscale systems
551 would allow for a sustained delivery of the biologics promoting long-term IVD regenerative processes.

552 The design of the optimal system for IVD regeneration is still under debate as disc degeneration is a
553 multifactorial disease. Cell-based therapies were widely considered in view of the major decrease in cell
554 numbers that is observed with degeneration. However, the regulatory framework and the costs associated
555 with these strategies may restrict their use. We believe that an effective approach would target the early
556 degenerative processes and involve multiscale biomaterials allowing molecules to be sustainably delivered
557 in situ and enabling a prolonged stimulation of endogenous repair.

558

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564

565 **9. Figures legends**

566

567 **Figure 1: Healthy intervertebral disc (IVD) local vascularization, innervation, and internal organization**

568 A. Located between vertebrae, the IVD is composed of two vertebral endplates, an *Annulus fibrosus*
569 (purple) and a *Nucleus pulposus* (NP, blue). Since the IVD is non-innervated and non-vascularized,
570 oxygen and glucose supply as well as metabolite excretion occur by diffusion through the vertebral
571 endplates. This particular property induces specific conditions of osmolarity and pH, leading the IVD
572 resident cells to adapt their metabolism. B. In AF tissue, fibroblast-like cells are organized parallel to
573 collagen concentric lamellae. In NP tissue, large vacuolated notochordal cells and nucleopulpoocytes
574 coexist and ensure homeostasis of the extracellular matrix.

575

576 **Figure 2: NP homeostasis rupture leads to IVD degeneration**

577 In a healthy IVD, regulation of anabolic and catabolic processes (i.e., homeostasis) enables the tissue to be
578 hydrated and maintains its mechanical properties. Under various factors such as genetics, aging,
579 environmental causes, or mechanical stress, the cellular dialogue underlying the extracellular matrix
580 homeostasis is disrupted. This rupture causes lower anabolism and increased catabolism, ultimately

581 leading to ECM degradation, tissue inflammation, vascularization, fibrosis, and nerve ingrowth.
582 Altogether, these phenomena are the key players in IVD degeneration.

583

584 Figure 3: Current IVD regenerative medicine is based on three approaches

585 **Cell-based therapies.** Considering the pivotal role of the decreased cell number in IVD degeneration,
586 cell-based therapies have been increasingly considered. Autologous or allogeneic cells have been
587 proposed and stem cells have been the subject of major investigations. Stem cells have been mainly
588 isolated from adipose tissue or bone marrow and more recently pluripotent stem cells have been
589 contemplated. It is also largely debated whether the optimal cells for IVD repair should be differentiated
590 or undifferentiated. **Molecular therapies.** The broadened understanding of the phenomena involved
591 during IVD degeneration has made it possible to consider molecular therapies via the direct injection of
592 various factors to promote anabolism, to manage inflammation, or to attract regenerative cells. **Nucleic**
593 **acid-based therapies.** Recent approaches target DNA and miRNA sequences and mostly aim to
594 downplay catabolic factor expression. Finally, combined approaches such as genetically modified cells
595 that could contribute to decreasing MMP local expression and activity could potentiate the efficacy of
596 these therapies.

597

598 Figure 4: Multiscale systems associate in situ-forming scaffolds with ready-to-use particles

599 In the context of IVD regenerative medicine, we propose to combine in situ-forming systems with ready-
600 to-use micro- or nanoparticles, thereby forming multiscale systems. We hypothesize that injectable
601 polymers that undergo local gel formation will restore IVD hydration and mechanical properties, while
602 micro- or nano-sized carriers loaded with cells or molecules will provide supportive cues for regenerative
603 processes.

604

605 Figure 5: Biomaterials: the driving force of the IVD regenerative medicine train

606 Biomaterials are essential for the design and development of delivery systems for IVD regenerative
607 medicine. They can protect and deliver nucleic acids, bioactive molecules, or cells for a long period and
608 may thus contribute to regenerating the degraded ECM of IVDs.

609

610

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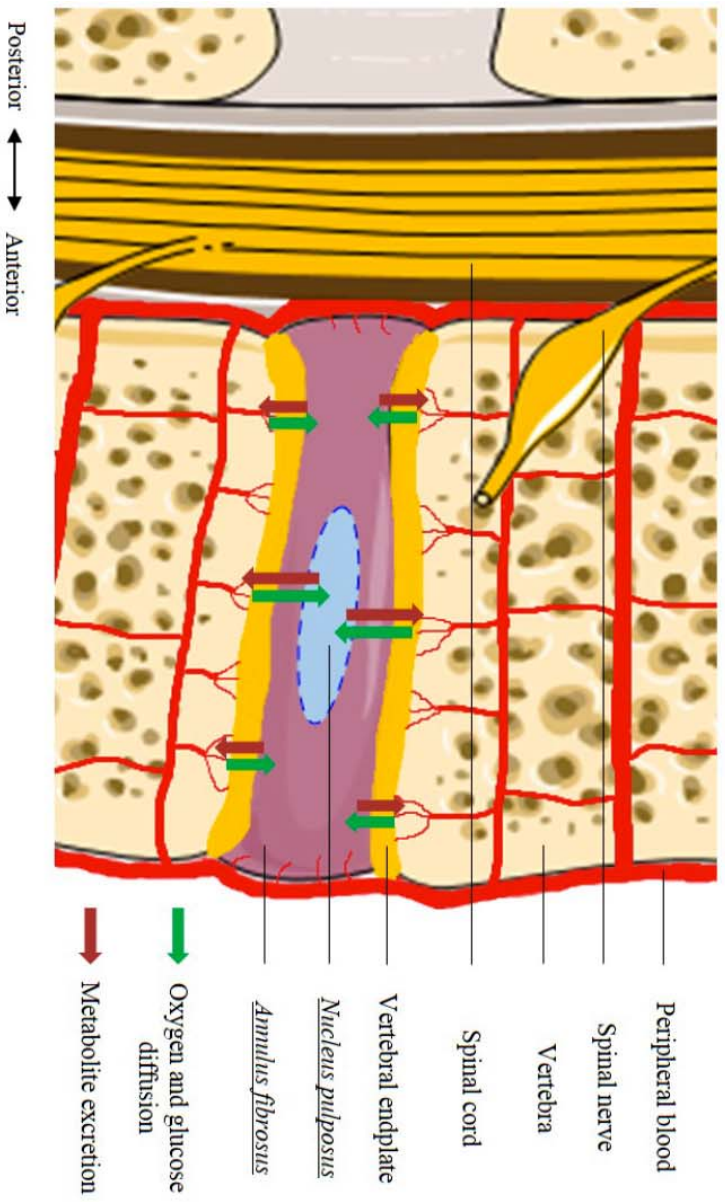
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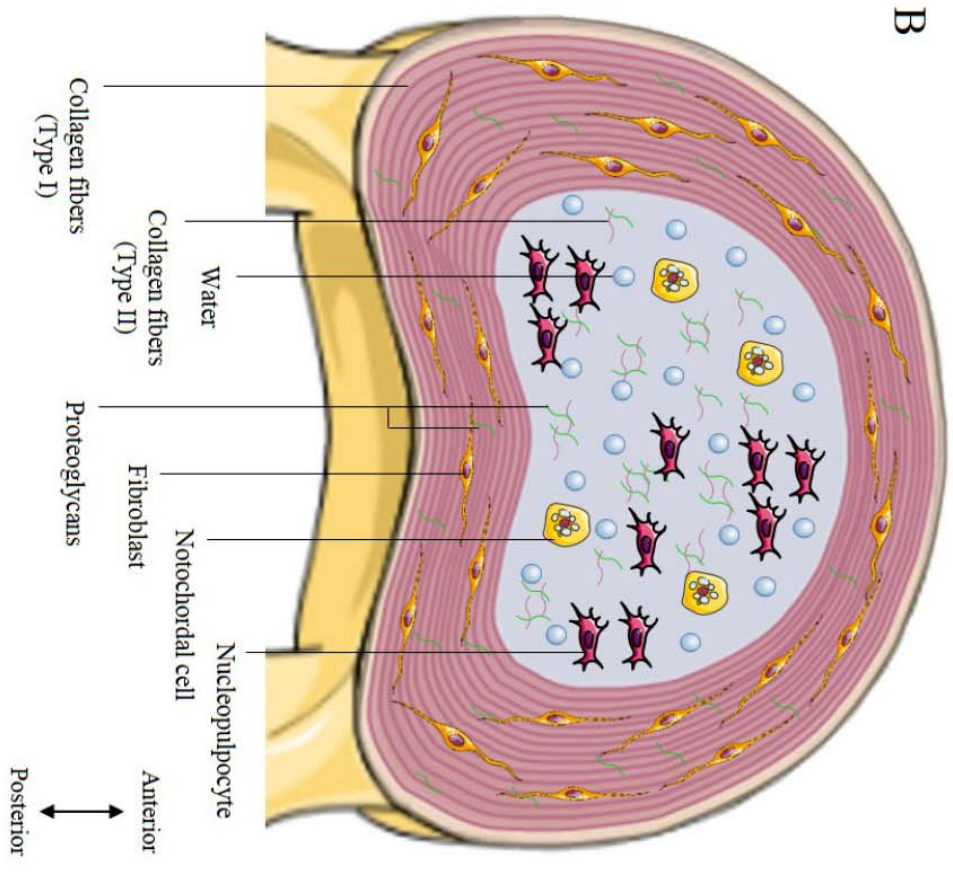
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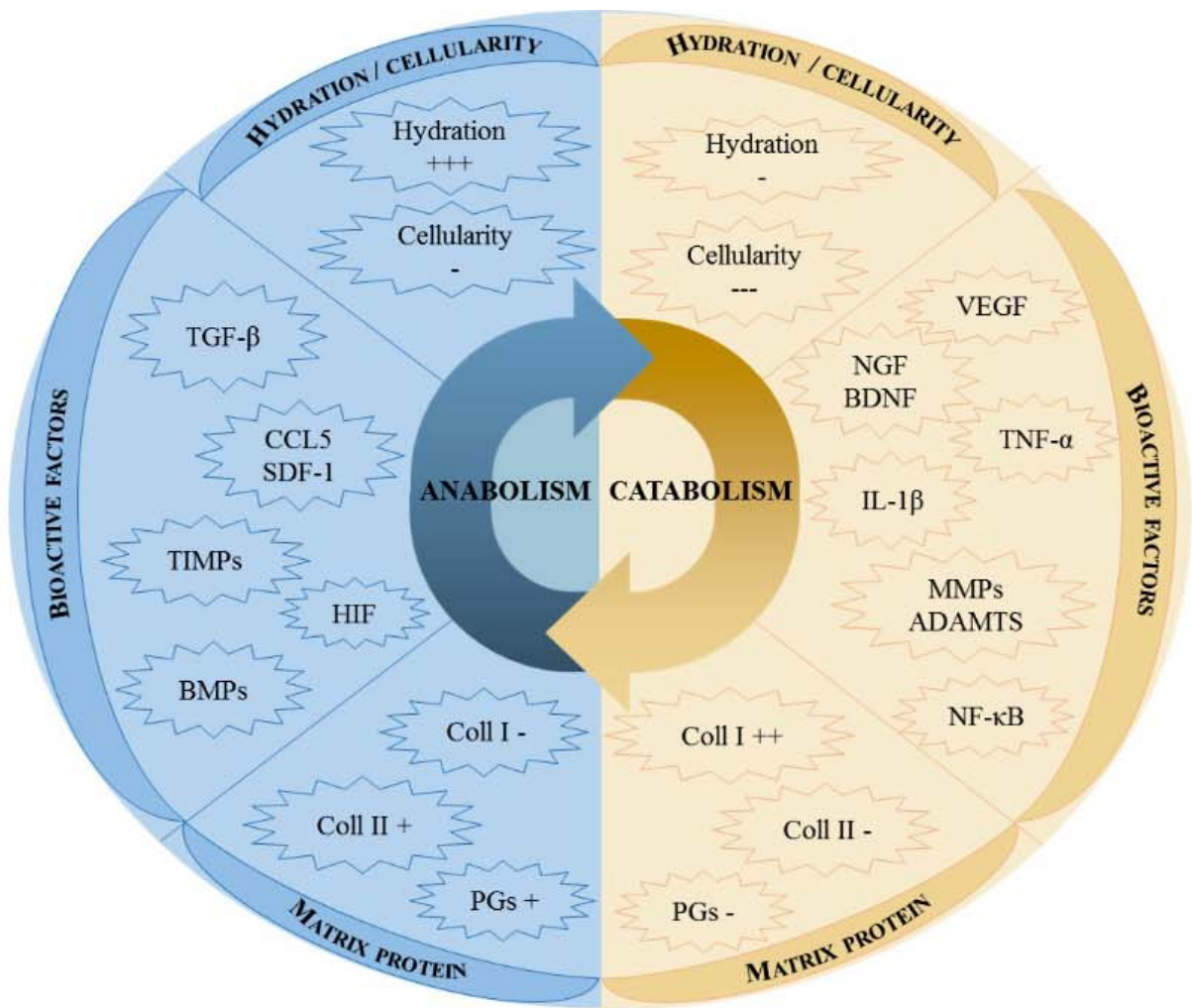
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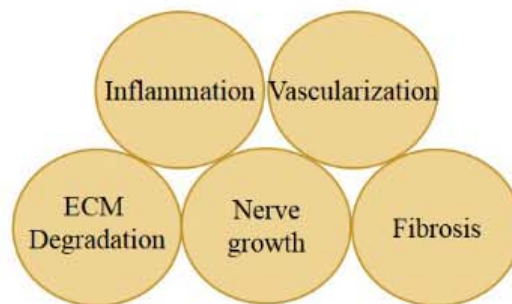


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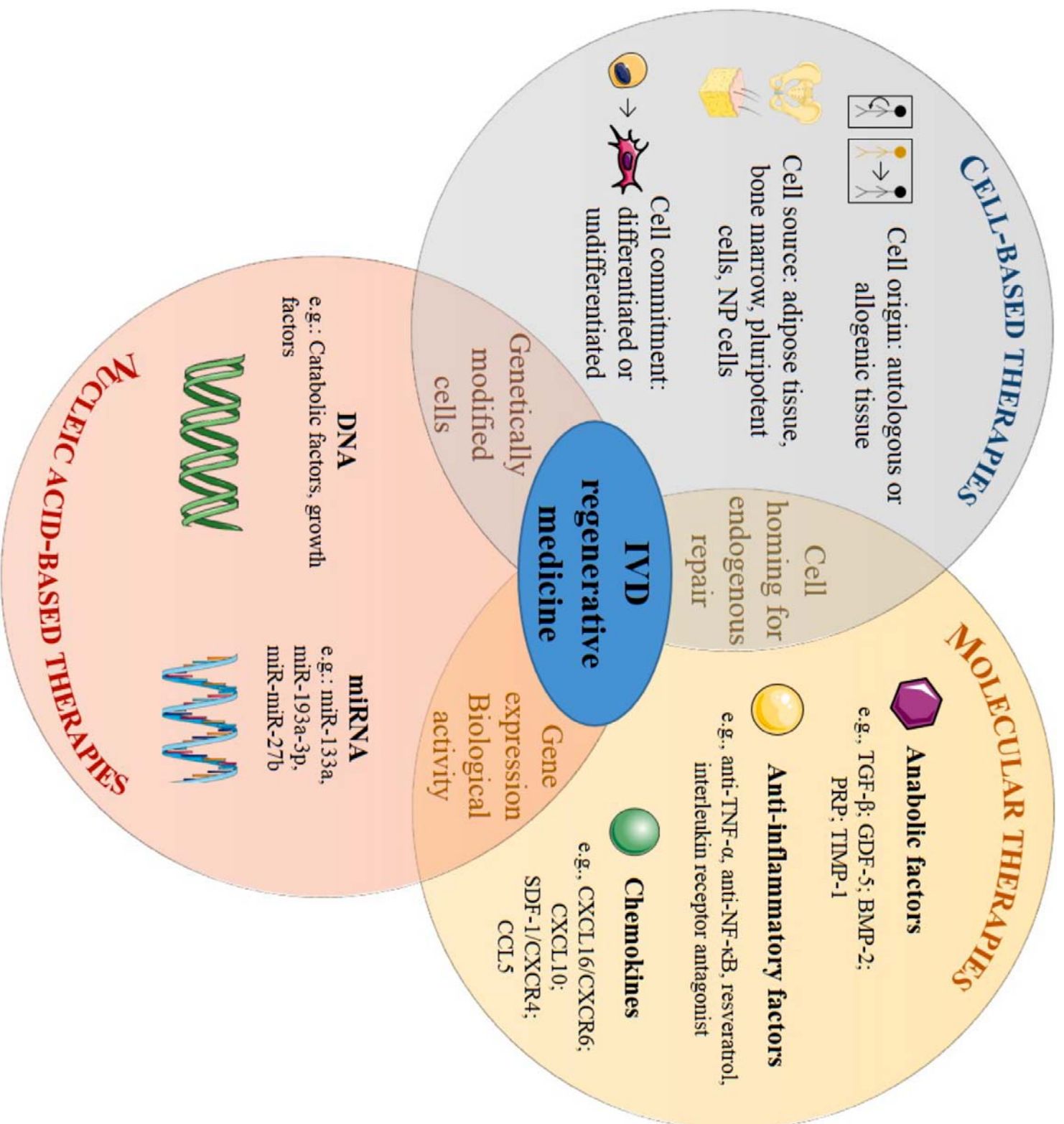




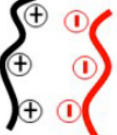
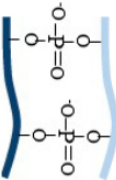





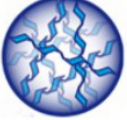
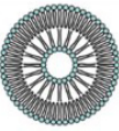
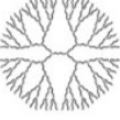




↓ Anabolism + Catabolism ↑



DISC DEGENERATIVE DISEASE



<i>In situ</i> forming systems	Ready-to-use systems	
<div style="display: flex; justify-content: space-between; align-items: center;"> <div style="width: 30%; text-align: center;"> <p>Macro-scaled scaffold</p> </div> <div style="width: 30%; text-align: center;"> <p>Micro-scaled particles</p> </div> <div style="width: 30%; text-align: center;"> <p>Nano-scaled particles</p> </div> </div>		
<p>Physical crosslinking</p> <div style="display: flex; justify-content: space-around;"> <div style="text-align: center;">  pH </div> <div style="text-align: center;">  Temperature </div> <div style="text-align: center;">  Ionic interactions </div> </div> <p>Chemical crosslinking</p> <div style="display: flex; justify-content: space-around;"> <div style="text-align: center;">  Crosslinker </div> <div style="text-align: center;">  Silanol groups </div> <div style="text-align: center;">  UV light </div> </div>	<div style="display: flex; flex-direction: column; align-items: center;"> <div style="margin-bottom: 20px;">  Microsphere </div> <div style="margin-bottom: 20px;">  Microcapsule </div> <div>  Porous microparticle </div> </div>	<div style="display: flex; flex-direction: column; align-items: center;"> <div style="margin-bottom: 20px;">  Matrix </div> <div style="margin-bottom: 20px;">  Liposome </div> <div style="margin-bottom: 20px;">  Dendrimer </div> <div style="margin-bottom: 20px;">  Carbon nanotubes </div> <div>  Silica nanofibers </div> </div>
<p>Injectable and <i>in situ</i> crosslinkable</p> <p>Low invasive approach</p> <p>Cell and molecule protection</p>	<p>Physicochemical properties easily modified for well-adapted sustained delivery</p>	<p>Suitable carriers for lipophilic drugs and gene delivery; mechanical reinforcement of hydrogel scaffold</p>

MULTISCALE SYSTEMS

