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# Cell dynamics and metabolism of the foreign body response: characterizing host-biomaterial interactions for next-generation medical implant biocompatibility

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## Abstract

Implantable medical devices (IMDs) collectively represent a critical mainstay in modern medicine. Used in many chronic diseases and in acute surgical interventions, IMDs are often associated with improvements in disease progression, quality of life, and mortality rates. Despite the positive impacts of IMD implementation, excessive fibrosis driven by the foreign body response (FBR) is frequently associated with the development of complications and failure. These complications in turn result in surgical revisions and removals, which represent a significant burden to healthcare costs and surgical wait-times in countries with elevated IMD usage rates. IMD complications are exacerbated by limitations to treatment options and limited availability of biocompatible materials. Novel treatment development is equally hampered by the complexity of the FBR, wherein complex cellular behaviors defy canonical immunological classification systems. In this review, current understandings of cellular dynamics and kinetics within the FBR are summarized, with a specific focus on the relationship between immunometabolic regulation and pathological fibrotic processes across various cell behaviours in the FBR. This review also explores promising emerging *in vitro* and *in vivo* techniques of FBR characterization, and highlights biomaterial properties associated with alterations in FBR outcomes. Finally, this review explores current and future approaches to biocompatible material development, highlighting immune-metabolic control as a therapeutic approach to mitigating the FBR.



# 1. The foreign body response to implanted materials: burden and clinical implications

Implantable medical devices (IMDs) have broad utility in the medical field, with applications spanning sutures, structural meshes, soft tissue fillers, orthopedic and craniofacial prosthetics, cerebral shunts, vascular stents, valvular prostheses, cardiac and neural stimulators, biosensors, contraceptive devices, and long-term drug eluting devices. For example, the single most common surgical procedure is the repair of primary or incisional ventral hernia, which is a prototypical example of soft tissue implant; current USA caseloads are approximately 611 000 per year and with a total cost of \$9.7 billion USD<sup>1</sup>. When combined with inguinal and femoral hernias, the incidence of medically significant cases remain undertreated, with combined incidence of approximately 13 million cases per year and representing a major source of morbidity, mortality, complications, and recurrence, as well as a significant component of healthcare spending<sup>2,3</sup>. With an aging population in the Western hemisphere and increased medical spending, along with continuous surgical innovation and the release of new IMDs, rates of surgical implantation are projected to increase for the foreseeable future.

Generally, IMDs positively impact recipient quality of life (e.g., hernia repair via a polypropylene surgical mesh), but also carry complication risks<sup>4</sup>. Specifically, IMDs are generally subject to some degree of chronic inflammation associated with the foreign body response (FBR) to implanted materials. The FBR is an expected inflammatory reaction, mediated by the immune system, that arises following implantation of a biomaterial within a host organism<sup>5</sup>, and that is characterized by the persistence of immune cells and the development of a fibrotic layer encapsulating and isolating the implant<sup>6,7</sup>. Based on patient, implant, and environmental factors, the FBR can manifest on a spectrum of severity. The FBR in a successful medical device application may present with no discernable impact on device function or patient quality of life. Conversely, a deleterious FBR can completely impair device function, severely incapacitate patients with pain or reduced quality of life, leads to systemic sequelae, or even threatens patient life altogether<sup>4</sup>. This pathology is characterized by a prolonged period of chronic inflammation, and the deposition of fibrotic matrix around the implant to isolate the area from prolonged tissue damage. This damage results from inflammatory molecules secreted by the immune system; generally, the more prolonged the inflammation resulting from implantation, the thicker the eventual fibrosis will be<sup>5</sup>. However, the immune system is also a powerful driver of tissue growth and healing, and in the future will likely be directly manipulated and harnessed in regenerative approaches<sup>8</sup>.

Despite the clinical ubiquity of IMDs and their respective complications, the mechanisms underlying the FBR are poorly understood. It is accepted that primary drivers of the FBR include chronic inflammation and the pathological recruitment of a variety of myeloid cells<sup>9</sup> (**Figure 1**). These cells drive excess fibrosis and limit tissue integration of the implant<sup>9</sup>. Typically, pathological inflammation and fibrosis manifests clinically as capsular contracture, resulting in pain and in extreme cases alteration of an implant's structure; symptoms may require surgical revision or explantation. Additionally, pathological fibrosis can interfere with implant function. For example, fibrotic depositions can alter drug elution rates in insulin delivery devices<sup>10</sup>, or implanted electrode function<sup>11–13</sup>. Though clear clinical complications are associated with pathological FBR fibrosis, the complement of involved cells, their activators and regulators, and their correlation to the final disposition of the FBR remain undercapitalized. More rare adverse outcomes can include the development of autoimmune diseases (e.g., breast implant illness) and even malignancy; these are generally thought to be result from prolonged local or systemic immunostimulation, but their study is hampered by little direct clinical study or model development<sup>14</sup>.

In this review, we discuss the mechanisms of the FBR, with focus on the dynamics and metabolism of participating cells in the peri-implant environment and in environments out of direct contact with systemic circulation. We review recent literature that elucidates the provenance of these cells, and the checkpoints that mediate inflammatory resolution and implant integration, or alternatively lead to chronic inflammation. Finally, we discuss novel scientific



45 advancements in the techniques that query patient explants and *in vivo* and *in vitro* models of the FBR, as well as  
46 metabolically-based solutions to the FBR and associated pathology.

## 47 2. Physiology of the foreign body response

48 The most harmonious explanation for the physiological utility of the FBR is twofold: the rapid repair of the wounded  
49 tissue including protection from the external environment, and the neutralization and sequestration of both active  
50 and latent pathogens introduced with the foreign body. The placement of an IMD generates a wound environment,  
51 including significant tissue damage, and thus drives signaling of canonical wound repair processes. In models of  
52 incisional skin wound healing (arguably the most studied of tissue wound healing processes), the first week of healing  
53 is characterized by a hemostatic plug that allows for inflammatory cell infiltration and sterilization, but imparts  
54 minimal tensile strength to the wound closure. This strength greatly increases in the second week, and progresses  
55 over the course of further weeks to months to a plateau of ~80% of normal dermis<sup>15,16</sup>. At a fundamental level, the  
56 FBR can be considered in the context of the wound healing paradigm, and the mechanistic drive for wound closure.

57 Early phases of wound healing utilize inflammatory pathways to eliminate damaged tissue, and address pathogens  
58 in the microenvironment. Pathogen exposure is a minimizable but unavoidable consequence of material  
59 implantation, and attributable to both endogenous (patient-colonizing) and exogenous sources (airborne particles  
60 in the operating room, films or particles from IMDs, surgical tools, and operators)<sup>17</sup>. Considering abdominal wall  
61 hernia repair meshes as an example, surgical site infection occurrence rates range enormously depending on wound  
62 and mesh size, mesh characteristics, surgical technique, environment, prophylaxis, dressing, and active wound  
63 management, as well as patient demographic factors<sup>17–25</sup>. The development of a fibrous barrier allows for spatial  
64 limitation of pathogen growth, as well as the protection of surrounding tissue from the cytotoxic anti-pathogen  
65 response from infiltrating immune cells during the FBR.

66 When considering the utility of implantable devices in surgical techniques, the FBR is critical for implanted materials  
67 to achieve stability in tissue and carry out their functional roles. The utility of controlled fibrosis increasing tissue  
68 hardness and stiffness is especially crucial in the face of recent medical devices. In the context of hernias, mesh  
69 repairs generally outperform non-mesh repairs in rate of primary recurrence<sup>26–28</sup>, generally attributable to the  
70 increased fibrosis at and around the surgical mesh; this suprphysiological stiffness of the mesh and hypertrophic  
71 scar tissue together allow for sufficient integration and tensile strength to ensure patency. Similarly, in silicone breast  
72 reconstruction, fibrosis is required for successful tissue integration and implant stability to minimize risk of  
73 displacement<sup>4,29,30</sup>. Without the FBR, implantable medical devices of all designations would be at risk for  
74 dislodgement and loss of function or even active pathology.

75 However, careful attention to detail is warranted in considering tissue and IMD mechanics. For example, the  
76 adaptation of polymer meshes originally designed for use in abdominal wall hernias for gynecological application in  
77 pelvic floor dysfunction, specifically in the adoption of transvaginal meshes, has been met with significant rates of  
78 complications<sup>31</sup>. These generally result from both differential elasticity under deformation between native tissue  
79 and the implanted material. As implantable meshes plastically deform, their interaction with softer surrounding  
80 native tissues often leads to erosion, which is often complicated by chronic pain, infection, dyspareunia, and  
81 autoexplantation or extrusion leading to perforation of surrounding organs, prolapse recurrence, or excess fibrosis  
82<sup>31</sup>. These complications occur much more frequently in the gynecological setting than the original abdominal wall  
83 setting, leading to widespread cancellation of medical device approvals by regulatory agencies, and the ongoing  
84 tempering of their recommendations by professional medical societies<sup>32–34</sup>. This differential outcome between the  
85 use of similar devices for different indications underscores the importance of application specific considerations (i.e.  
86 microenvironmental signaling and mechanics) in designing surgical solutions. To understand the factors underlying  
87 these critical mechanics of the host-wound interface, a thorough understanding of relevant biology is warranted.

88



## 89 2.1. Timeline of the foreign body response

90 Placement of an IMD induces a wound healing response, wherein the device is placed in an environment of tissue  
91 damage. After the initial abrasion is created, and the implant is introduced into the body, an *adsorption period* begins  
92 <sup>5</sup>, during which host proteins, extracellular matrix (ECM) and cell debris adsorb and desorb fluidly at the surface of  
93 the implant <sup>7</sup>. Due to the tissue damage associated with implantation, adsorption occurs effectively immediately  
94 (i.e., reaching adsorbed homeostasis under 30 min) to an implant surface <sup>35</sup>. The damaged tissues and denatured  
95 proteins surrounding the implant serve as damage-associated molecular patterns (DAMPs), which engage the cells  
96 of the immune system through pattern recognition receptors (PRRs) to respond to tissue damage <sup>36</sup>. Common DAMPs  
97 include ECM proteins such as fibrinogens, fibronectin and heparan sulfate, all of which initiate immune signaling  
98 through toll like receptor (TLR) <sup>4</sup> <sup>37–39</sup> as well as intracellular components (DNA, RNA, histones, etc.) <sup>40</sup> and adsorbed  
99 immune proteins such as complement elements and antibodies <sup>7</sup>. Proteins adsorbed at the surface of the implant  
100 promote the recruitment of platelets, induction of complement cascading, and immune cell chemotaxis through  
101 cytokine signalling <sup>36</sup>. Receptor-binding to TLRs underlies early inflammation through induction of inflammatory  
102 mediators such as IL-6 and interferons, and appropriately each TLR recognizes a finite number of ligands. Differential  
103 conformation and adsorption profiles of DAMPs may be an important factor in the inflammatory profiles associated  
104 with different materials <sup>41</sup>. In the FBR, adsorbed DAMPs (namely fibronectin and fibrinogen) have implicated both  
105 TLR2 & TLR4 as activating PRRs <sup>6,42</sup> with downstream activation of NF- $\kappa$ B and pro-inflammatory mediators. However,  
106 the breadth of DAMP signalling engaged during the FBR likely involves other endogenous ligand-binding TLRs (e.g.,  
107 TLR3, TLR7 and TLR9) and C-type lectin receptors <sup>36</sup>. Ultimately, more research is required to identify the entirety of  
108 PRRs engaged during the FBR.

109 As soon as biochemical adsorption has occurred, cell reaction local to the foreign body induces resident immune cell  
110 activation and recruitment of more distal or systemically circulating immune cells. These events constitute the *acute*  
111 *inflammatory phase* of the FBR, lasting for hours to days <sup>5</sup> and characterized by nonspecific cell activity optimized  
112 for rapid neutralization of invading organisms and systemic immune escalation and recruitment (**Table 1**).  
113 Neutrophils are the ‘first responders’ of the FBR and secrete pro-inflammatory cytokines <sup>5,7,36</sup>. Neutrophil  
114 extracellular trap (NET) activity, reactive oxygen species (ROS) production and chemoattractant production are also  
115 induced by neutrophils <sup>5,7,36</sup>. To date, the balance of evidence suggests key roles of these functions in <sup>43–45</sup> in  
116 propagating deleterious FBRs, although differences may be minimal or only manifest later in the course of healing  
117 and fibrosis <sup>43–45,46</sup>. Additionally, mast cells have long been recognized as critical to the early FBR, as well as to the  
118 fibrotic process; early mast cell degranulation is ostensibly responsible for neutrophil and monocyte extravasation  
119 and chemotaxis to the peri-implant environment in the context of the FBR <sup>47,48</sup>. The mast cell response, although  
120 somewhat lacking in recent study, has been linked specifically to near-instantaneous fibrinogen binding and  
121 activation on the surface of implanted biomaterials <sup>49</sup>, with modulation of this pathway robustly affecting the early  
122 response with a negligible to moderate effect on fibrotic outcome <sup>38,47,48</sup>. Despite long-standing knowledge,  
123 mechanistic study of mast cell contributions to the FBR have been hampered by challenges associated with *in vitro*  
124 study and in modulating the mast cell response experimentally *in vivo* <sup>50</sup>.

125 The release of inflammatory mediators during the acute inflammatory phase contributes to the recruitment of  
126 macrophages, critically involved in the FBR <sup>5,7,36</sup>; the continued recruitment of these cells and persistence beyond  
127 that of mast cells and neutrophils, and forms the *chronic phase* as specific to the field of biomaterials and the FBR  
128 <sup>5,51</sup>. Macrophages, in turn, respond to DAMPs associated with the implant surface through secretion of inflammatory  
129 molecules <sup>5,7,36,52</sup>. During normal wound healing processes, inflammatory macrophages are short-lived. This is in  
130 contrast with macrophages in the FBR, which remain inflammatory due to the persistence of the implant and  
131 denatured proteins within the host. In the absence of foreign material, this chronic phase gives way to a resolution  
132 phase; continued exposure to foreign material instead induces granulation in the *FBGC formation phase*, which will  
133 be discussed below.



134 To protect the body from the harmful products being released by immune cells that direct the chronic inflammatory  
135 response at the implant interface, macrophages secrete pro-fibrotic cytokines, and recruit fibroblasts which deposit  
136 a collagenous layer of tissue around the implant<sup>5,7,36,52</sup>. This collagenous layer is referred to as the fibrotic capsule.  
137 The fibrotic capsule plays a critical role in determining the negative outcomes of the FBR, including discomfort, pain,  
138 capsular contracture and implant failure<sup>4</sup>. This process of fibrotic encapsulation and ongoing immune reaction at  
139 the implant-interface constitute the final phase of the FBR, the *fibrous capsule* phase<sup>5,7,36,52</sup>. As the capsule continues  
140 to form, a variety of pro-inflammatory macrophage- and FBGC-secreted cytokines, in particular vascular endothelial  
141 growth factor (VEGF), mediate angiogenesis and recruitment/activation of myofibroblastic processes<sup>9</sup>. Finally,  
142 although the formation of a foreign body granuloma does not require adaptive immunity<sup>53</sup>, adaptive immune cells  
143 (i.e., lymphocytes) are present in large quantity in local region<sup>43,54,55</sup>. There is currently little evidence to support  
144 direct antigenicity of synthetic polymeric biomaterials, but FBRs have featured T- and B-cells interacting either  
145 directly with biomaterials or in response to macrophage secretions, and then acting as intermediaries and amplifiers  
146 to cytokines and chemokines by other cells<sup>54,56–59</sup>.

147

## 148 2.2. Macrophage phenotypes associated with the foreign body 149 response

### 150 2.2.1. Canonical macrophage classification systems

151 The canonical classification system for macrophages is the M1/M2 paradigm, wherein M1 macrophages are  
152 considered pro-inflammatory or classically activated, and M2 macrophages are considered pro-reparative or  
153 alternatively activated<sup>60,61</sup>. The role of M1 vs. M2 macrophages in directing inflammation and its resolution has been  
154 excellently profiled elsewhere<sup>62</sup>; classical activation generally results from stimulation with lipopolysaccharide (LPS)  
155 and interferon (IFN) $\gamma$ , which is associated with pathogen and damage clearance<sup>63</sup>. Phenotypically, M1 macrophages  
156 are characterized by elevated MHC-II and CD-80/CD-86 expression, indicative of antigen presentation and co-  
157 stimulatory T-cell activation respectively<sup>63</sup>. Generally, alternative activation results from stimulation with reparative  
158 cytokines such as IL-4, IL-13 and IL-10, and M2 macrophages are involved in tissue remodelling and debris clearance  
159<sup>63,64</sup>. Phenotypically, M2 macrophages are generally identified as expressing CD-206 (or the macrophage mannose  
160 receptor)<sup>63</sup>, however the M2 grouping of macrophages is diverse. A plethora of stimulating factors inducing the  
161 expression of a variety of phenotypic markers has resulted in the generation of subgroups for specification of cellular  
162 behaviors known collectively as M2.

163 There is an increasingly pervasive skepticism towards this classification system however, particularly in the context  
164 of complex environments like the FBR, as focused investigation reveals that macrophage phenotypes have indicated  
165 a greater number of exceptions to the M1/M2 rule than not<sup>60,61,65</sup>. In the tissue-implant microenvironment,  
166 macrophages show elements of both M1 and M2, due to their involvement in inflammation and remodelling<sup>65</sup>; this  
167 may help to explain the generally-accepted observation that a “runaway” M2 response is considered to be a  
168 contributor to the pro-fibrotic component of the FBR. Macrophages may first require inflammatory activation to  
169 develop a robust M2-like response, which may contribute to this phenotype in the chronic inflammatory  
170 environment of fibrosis<sup>66</sup>. Oversimplification of macrophage subsets is further complicated by heterogeneity in  
171 phenotypes in the fibrotic tissue microenvironments; macrophages of both characteristic phenotypes that may be  
172 arranged in a spatially distinct fashion, contributing to the macroscopic phenotype of the environment<sup>67</sup>.

173 Further nuance can be attributed to tissue-specific macrophage metabolic preferences in inflammation<sup>68</sup>, findings  
174 of systemic immunity or its interactors (e.g. gut microbiome) influencing the response to implanted materials<sup>69</sup>, and  
175 findings that various macrophage functions (e.g., secretion of individual cytokines) seem dependent on different  
176 metabolic pathways and again differential per activation profile<sup>70</sup>. Efforts to establish an FBR-specific macrophage





177 phenotype are progressing, although clear models of function remain challenged by discrepancies between different  
178 experimental models and techniques <sup>71</sup>. In the FBR, macrophages produce pro-inflammatory cytokines <sup>72</sup> such as  
179 Tumor Necrosis Factor (TNF), interleukin-6 (IL-6), and interleukin-1 $\beta$  (IL-1 $\beta$ ) as well as pro-reparative cytokines such  
180 as Transforming Growth Factor  $\beta$  (TGF $\beta$ ), leading to speculation that macrophages in this environment might  
181 represent a novel phenotype, ill-defined by the M1/M2 system <sup>60,61,65</sup>, but characterized by an environment rich in  
182 IL-4, IL-17, and IL-34 <sup>59,73</sup>.

183

## 184 2.2.2. Correlating function to metabolic macrophage phenotypes

185 In service of better defining macrophage functional phenotype, increased effort has been focused on understanding  
186 the metabolic foci that enable domains of macrophage activity <sup>74</sup>. The general relationship between macrophage  
187 cellular function and metabolic behavior is well established: typically, 'inflammatory' macrophages rely on glycolysis  
188 and 'reparative' macrophages rely on oxidative phosphorylation <sup>75</sup>. Glycolytic macrophages in the early wound  
189 environment are marked by upregulated *HIF1 $\alpha$*  expression and stabilization. These cells are enriched for *SLC2A1*  
190 (encoding GLUT1), as well as the expression of pro-inflammatory and pro-angiogenic genes, implicating the  
191 functionality of early wound macrophages in organizing vasculature and wound sterilization <sup>76</sup>. These differences are  
192 functional: at the most surface level, inflammatory macrophages tend towards glycolysis as it facilitates the rapid  
193 production of ATP, without the taxing need for mitochondrial biogenesis. However, the factors linking glycolysis and  
194 inflammation are more intricate. M1 macrophages result from stimulation with LPS and IFN- $\gamma$ . LPS is an inducer of  
195 the NOX2 pathway, which in turn yields NADPH, critically utilized by phagocytic cells to meet the reactive oxygen  
196 species (ROS) requirement of the phagosome. Induction of this pathway has been demonstrated to require  
197 stabilization of HIF1 $\alpha$  by succinate accumulation, itself enabled by the TCA cycle arrest characteristic of glycolytic  
198 reprogramming. The glycolytic inhibitor 2-deoxyglucose (2-DG) has been known to negatively impact induction of  
199 the NOX2 pathway and the associated production of NADPH in LPS-stimulated macrophages <sup>77</sup>. HIF1 $\alpha$  more generally  
200 serves as a transcriptional regulator of a variety of inflammatory and glycolytic genes, serving as the best example  
201 of the overlap between glycolytic programming and inflammatory gene induction. HIF1 $\alpha$  suppression has been  
202 demonstrated to reduce NLRP3 inflammasome activity and IL-1 $\beta$  signaling in alveolar cells following bleomycin-  
203 induced lung injury, linking inflammation and glycolysis, specifically within in the context of NLRP3 and wound-  
204 healing <sup>78</sup>. The NLRP3 inflammasome is closely linked to aggressive anti-pathogen functionality in macrophages <sup>79,80</sup>,  
205 and in the context of the FBR its inhibition can attenuate both inflammation and fibrosis <sup>72</sup>. A growing body of  
206 evidence suggests that NLRP3 is metabolically regulated through glycolytic metabolites, TCA intermediates  
207 (including succinate, itaconate, and fumarate), and both saturated and unsaturated fatty acids <sup>79–81</sup>. However, the  
208 precise mechanisms of these regulatory controls remain ambiguous due to differences between studies in knockout  
209 design, *in vitro* vs. *in vivo* experimentation, cell type, and the inherent pleiotropic roles of many such metabolites.

210 In contrast, 'reparative' (i.e., M2) late stage wound healing macrophages are putatively characterized by intact TCA  
211 cycles (OXPHOS metabolism), associated with reduced ROS production <sup>82</sup>. These cells can utilize glycolysis to  
212 generate succinate as fuel for the TCA cycle, but have demonstrated the ability to maintain TCA cycle usage during  
213 glycolytic inhibition, likely through increased reliance on fatty acid oxidation (FAO) <sup>83</sup>. These changes are once again  
214 functional. Reparative macrophages are not as reliant on rapid energy turnover and the availability of synthetic  
215 intermediates, and as such do not require upregulated glycolysis and PPP to the same degree as inflammatory cells.  
216 FAO, which is notably increased in 'M2' macrophages, is typically associated with longer cellular lifespans, which  
217 would be useful in orchestrating prolonged periods of tissue remodeling.

218 Additionally, reparative signals and oxidative phosphorylation are linked. Various studies have pointed to the  
219 reduction in OXPHOS following IL-10 suppression, pointing to IL-10 as having a role in metabolic orchestration.  
220 Induction of these metabolic pathways are believed to contribute to the synthesis of wound-healing intermediates  
221 such as collagen <sup>84</sup>. IL-10 supports successful wound healing, and has been documented as anti-fibrotic by acting on



222 myofibroblasts to reduce collagen gene expression and lower  $\alpha$ -SMA production<sup>85</sup>. Conversely, other studies have  
223 indicated reductions in IL-10 signaling following OXPHOS interruption, suggesting that there is reciprocity to the  
224 regulation of IL-10 on OXPHOS<sup>86</sup>. Mechanistically, these pathways are still unclear. Further, the precise metabolic  
225 distinctions between M2 subgroups remain ambiguous. For example, the M2b subgroup of macrophages are  
226 implicated in “M2-mediated inflammation”, and as such it is unlikely that these cells rely on anti-inflammatory  
227 OXPHOS and FAO pathways to the same extent as M2a macrophages, which are more strongly implicated in tissue  
228 remodeling. A 2023 publication demonstrated the capacity for strong HIF1 $\alpha$  associated glycolytic reprogramming of  
229 dually IL-4- and IL-13- stimulated macrophages. Though these cells showed glycolytic upregulation and succinate  
230 accumulation characteristic of M1 metabolic programming, they also maintained certain metabolic and phenotypic  
231 features characteristic of the M2 lineage, namely arginase 1 (Arg1) activity<sup>87</sup>. This study is useful in illustrating the  
232 potential incongruence between classification by metabolic behavior and canonical phenotyping.

233 Critically, within the context of the FBR, the phenotypic characteristics of macrophages are ill-defined, and reliable  
234 information regarding the metabolic behavior of these cells is even more evasive. Thus, any inferences with regards  
235 to the metabolic behavior of macrophages during the FBR are speculative and require further demonstration.  
236 However, the engagement of key cellular pathways during the FBR may shed light on the metabolic behavior of  
237 macrophages during this reaction. For example, NLRP3 inhibition has been observed to reduce implant-associated  
238 fibrosis following nerve-injury in mice<sup>72</sup>. The links between NLRP3 inflammation, IL-1 $\beta$  expression, and glycolysis  
239 would implicate glycolysis in the chronic inflammation associated with implants. The role of glycolysis in the FBR is  
240 also supported by the high-flux HIF-1 $\alpha$ -mediated glycolytic metabolism observed in early wound-healing  
241 macrophages, though the persistence of HIF-1 $\alpha$  signaling in chronic FBR tissues remains to be demonstrated<sup>37,76</sup>.

242 Though the induction of TGF- $\beta$  signaling by implant-associated macrophages is universally believed to play a key role  
243 in the pathology of the FBR<sup>5</sup>, the cellular populations responsible for excessive TGF- $\beta$  signaling in the FBR remain  
244 unclear<sup>88</sup>. This is potentially once again attributable to the cells in the FBR displaying phenotypic plasticity, with  
245 ‘M1’-like cells and ‘M2’-like cells being equally implicated in TGF- $\beta$  signaling. Interestingly, despite being largely  
246 attributed to ‘M2’-like cells (which are primarily oxidative), HIF-1 $\alpha$  upregulation has been shown to induce TGF- $\beta$   
247 production by human macrophages<sup>89</sup>, and specifically has a documented supporting role in TGF- $\beta$  mediated  
248 transcription of fibrotic genes by alveolar macrophages during bleomycin-induced lung fibrosis in mice<sup>90</sup>. This could  
249 imply a relationship between hypoxia, glycolysis and fibrosis, though metabolic differences between the  
250 macrophage phenotypes studied would require confirmation in the FBR context. Adding further complexity, IL-10  
251 has been observed to reduce TGF- $\beta$  production by alveolar macrophages, and ultimately reduce bleomycin-induced  
252 lung fibrosis, despite the commonly held belief that both of these cytokines stem from the same cellular origins<sup>91</sup>.  
253 Ultimately, further analyses are required to determine the links between metabolism and excessive TGF- $\beta$  signaling.

254 Finally, classification of macrophages in the FBR is further impaired by their formation of foreign body giant cells  
255 (FBGCs). FBGCs are large, terminally differentiated, multinucleated cells with poor phagocytic, and enhanced  
256 lysosomal abilities, suggesting a role in debris clearance and extracellular degradation; they are the hallmark of the  
257 FBR, and persist at the tissue-implant interface for the remainder of the implant life cycle<sup>5,7,36,52</sup>. Multiple studies  
258 have demonstrated functional roles for IL-4 and IL-13 in macrophage fusion and FBGC formation through the  
259 upregulation of mannose receptors<sup>7,92,93</sup>, but the exact mechanisms and conditions through which macrophage  
260 fusion occurs in the FBR are still relatively unclear. The functional role of FBGCs, however, primarily involves  
261 extracellular degradation and phagocytosis, two processes which are strongly linked to glycolysis through HIF1 $\alpha$ ,  
262 though this link requires further study.

263





### 2.3. Foreign body giant cells

264  
265 The hallmark histologic feature of the FBR is the persistent presence of both macrophages and the multinucleated  
266 giant cell (MNGC) or FBGC, the latter of which are optimal for the phagocytosis and breakdown of large particles  
267 (e.g., 45  $\mu\text{m}$  or even larger<sup>94</sup>) that eclipse the capability of macrophages. There are multiple types of MNGC in the  
268 human body, which can be differentiated both by mechanism of formation as well as function or profiling. The best-  
269 characterized MNGs are arguably osteoclasts (mediating physiological bone remodeling in pathogen-free local  
270 environments) and Langhans giant cells (mediating granuloma formation in the presence of persistent and  
271 recalcitrant microbes such as *M. tuberculosis* and *M. leprae* as well as Bacillus Calmette-Guérin (BCG) vaccine)<sup>95</sup>.  
272 FBGCs are poorly-characterized in contrast to these other cell types, but have been positively identified due to  
273 different cytoplasmic structure and function as distinct from both osteoclast and infectious granulomatous MNGCs  
274<sup>53</sup>. FBGCs have emerged as key mediators of the acute and chronic FBR with wide-ranging sensing and activity<sup>96</sup>,  
275 including demonstrated roles in extracellular degradation via protease excretion, large particle phagocytosis,  
276 chemokine and cytokine release for immunomodulation, and antigen presentation<sup>6,7,95,97–100</sup>.

277 Despite longstanding awareness of the existence of FBGCs, little is known about both their formation and their  
278 resolution. Consensus from both *in vivo* and *in vitro* study, including *in vitro* fusion protocols and microscopic and  
279 molecular evidence, strongly supports the fusion of macrophages as the major factor in MNGC formation<sup>94,99–103</sup>.  
280 However, a recent study suggested that pattern recognition receptor-induced polyploidy and frustrated mitosis were  
281 the major drivers of MNGC formation in a model of BCG granulomatosis<sup>104</sup>; these mechanisms would merit further  
282 investigation in a model of sterile granuloma such as the FBR. In the case of material implantation specifically, foreign  
283 body MNGCs are induced by IL-4 and IL-13<sup>94,99,103</sup> secreted ostensibly from a non-T-cell source<sup>105</sup> and aspects of this  
284 activity have been recapitulated *in vitro*<sup>42,106</sup>, although the precise mechanism at play and co-stimulators *in vivo* are  
285 unclear<sup>96</sup>. It is equally noteworthy that while mast cells are well-documented producers of IL-4 and IL-13<sup>107</sup>,  
286 macrophage fusion has been documented following implantation in mast-cell deficient mice, further complicating  
287 the understanding of fusion signals resulting in FBGC formation<sup>108</sup>.

288 Biomechanical signaling has also been implicated via various integrins<sup>7,100,102,109</sup> and TRPV4-sensed stiffness<sup>110</sup> in  
289 combination with soluble cytokines<sup>111</sup>, which represent potential control points for improving implant  
290 biocompatibility. Additionally, TLR-mediated detection of adsorbed self proteins (i.e., DAMPs)<sup>7,42,112,113</sup> is correlated  
291 with GC formation or function. Again in the context of an infectious granuloma, complement opsonization (e.g., C3)  
292 is a potent trigger for MNGC function<sup>94</sup>; FBGCs may retain this activity, which could even be elicited in the context  
293 of non-opsonization sterile complement adsorption to protein-naïve biomaterial surfaces as part of the protein  
294 corona during the tissue damage associated with implantation, and so alter the course of the FBR and the material-  
295 specific safety profile<sup>7,100,114</sup>. T-cell contributions to FBR development and MNGC formation have been noted in  
296 both infectious (e.g., tuberculosis and parasite) and autoimmune pathology<sup>53</sup>. T-cell antigen response may be  
297 observed in certain instances of foreign body introduction, such as chronic beryllium disease, but other reactions  
298 such as silicosis have not yet demonstrated antigen-directed T-cell response beyond nonspecific pro- and anti-  
299 inflammatory helper T-cell and regulatory T-cell involvement<sup>53</sup>.

300 Many MNGCs display aspects of both M1 and M2 between function and metabolic phenotype<sup>115</sup>. One mechanism  
301 might be macrophage plasticity, in which classically-activated (“M1”) macrophages become resistant to TLR signaling  
302 and limit their pro-inflammatory activity while maintaining anti-inflammatory IL-10 release<sup>100</sup>. This complements  
303 both the observation that FBGCs develop after high proportions of M1 cells in the early FBR<sup>42</sup>, and the hypothesis  
304 that pro-inflammatory macrophages may take on anti-inflammatory properties and fuse upon stimulation with IL-4,  
305 IL-13 or vitamin E or tocopherol *in vivo*<sup>103</sup>. Here, the TLR2 pathway frustration may be synergistic or even priming to  
306 cytokine-mediated transdifferentiation and cell fusion<sup>116</sup>. Fibrinogen dependency of FBGC occurrence has been  
307 noted<sup>38</sup>, which correlates to mast cell degranulation during introduction of the material<sup>48</sup>. Material properties have  
308 long been known to influence FBGC activity, density per unit area, and ploidy<sup>117</sup>. Given the context dependence on  
309 the varied functionalities of FBGCs (including cytokine secretion, phagocytosis, and enzyme secretion), it is likely



310 difficult to directly correlate FBGC frequency *in situ* to the ultimate outcome of an implant across conditions. Instead,  
311 the trajectory of cell function likely remains the gold standard by which to assess the relevance of FBGC presence to  
312 FBR course.

313

## 314 2.4. Clinical models of macrophage and giant cell function and 315 dysfunction: the placenta and infectious vs. non-infectious granulation

316 Complexities of correlating histological occurrence to tissue function notwithstanding, macrophage and GC presence  
317 have long been correlated to chronic inflammation and disease outcomes based on their persistence and density.  
318 Given their often deleterious roles in pathological FBR occurrences, it is understandable that there is considerable  
319 effort to better understand the factors that mechanistically underly GC occurrence and resolution to allow for design  
320 of materials that minimize these factors of the FBR and thus improve clinical outcomes <sup>113,118,119</sup>. Therefore,  
321 additional insight in GC formation, persistence, and resolution may be found in the physiological and pathological  
322 models of the placenta and sarcoidosis, respectively.

323 The placenta is a high-throughput fetal-maternal interface for the exchange of dissolved O<sub>2</sub>, nutrients, and waste,  
324 forming from the embryonic trophoblast. It primarily shields the mother from the fetus to prevent development of  
325 anti-fetus immunity, although it also restricts certain antibody classes and cell-mediated immunity. This shielding  
326 often breaks down in the presence of previous RhD antigen alloimmunization, but generally is of exceptionally high  
327 quality. Of note, the placenta features a high concentration of GCs <sup>120</sup>. As in the FBR, M1 and M2 distinctions break  
328 down in the placenta. There appears to be a time course of dominant signatures, but there remain both M1 and M2  
329 markers throughout the course of normal pregnancy, and both molecular signatures and function of both M1 and  
330 all M2 subtypes persist in the healthy placenta across all stages of pregnancy <sup>120,121</sup>. M1 markers are more strongly  
331 expressed in the first trimester and aid in implantation before M2 markers demonstrate a receptivity to trophoblast  
332 invasion, vascularization, and placental development <sup>121</sup>. Later in pregnancy, high M1 activity is correlated with  
333 preeclampsia <sup>120,122</sup>, drawing similarities to sustained inflammation and systemic sequelae in sterile FBR-related  
334 illness.

335 In response to infectious insults, granulation can often be effective at indefinitely sequestering an agent, or even  
336 eliminating it completely and resulting with a sterile granuloma. In contrast, these reactions as well as sterile  
337 autoimmune reactions such as Crohn disease, sarcoidosis, vasculitides, or various hypersensitivities manifesting in  
338 granulation can progress to highly deleterious impact even based solely on mass effect and resulting tissue  
339 dysfunction <sup>53</sup>, separate from any systemic effect of chronic immune activation. In many cases, the provoking antigen  
340 or insult is never found. Further understanding of the induction, propagation, and growth of granulation processes  
341 may be highly valuable to clinical resolution of these sterile diseases as well as the FBR.

342

## 343 2.5. Macrophage direction of myofibroblast activity

344 Myofibroblasts are most responsible for maladaptive fibrotic deposition during FBR. Though direct TGF $\beta$  signaling  
345 from macrophages is known to promote myofibroblast induction, macrophages participate in promoting fibrosis  
346 through many pathways <sup>123,124</sup>. Early wound macrophages release IL-6, which induces paracrine TGF $\beta$  signaling in  
347 fibroblasts, and promotes differentiation into myofibroblasts <sup>125</sup>. IL-6 can also induce myofibroblasts directly, by  
348 promoting  $\alpha$ -SMA via a JAK1-ERK signaling pathway <sup>126</sup>. PDGF, another well-documented macrophage-associated  
349 inducer of fibrosis, promotes myofibroblast activation seemingly through promoting paracrine TGF- $\beta$  signaling <sup>127</sup>.  
350 Finally, macrophages can seemingly induce myofibroblast activity through TGF- $\beta$  independent mechanisms.  
351 Specifically, RELM $\alpha$  expression induces  $\alpha$ -SMA expression in fibroblasts through Notch1 signaling <sup>128</sup>. TNF has also  
352 demonstrated pro-fibrotic roles in several pathologies, through the induction of collagen synthesis, proliferation and



353 activation of myofibroblasts<sup>129,130</sup>. IL-1 $\beta$  further drives fibrotic outcomes through indirect promotion of IL-6, TGF- $\beta$   
354 and PDGF<sup>131</sup>, although the mechanism is less obvious and warrants further study<sup>132</sup>. FGF-2, TGF $\beta$ , TNF $\alpha$ , PDGF as  
355 secreted by macrophages and FBGCs have been associated with myofibroblastic activity and wound/implant fibrosis  
356 and fibrotic disease pathology<sup>11,37,114,133</sup>. Heightened local VEGF levels have been associated with both increased<sup>9</sup>  
357 and decreased<sup>134,135</sup> fibrosis. Local ischemia can induce VEGF secretion, which can both support increasing capsular  
358 formation, or stabilize healthy healing interstitium; VEGF control can therefore potentiate both pro-fibrotic and pro-  
359 regenerative FBRs based on other physical, chemical, and temporal cues<sup>136</sup>.

360 The interaction between macrophages and fibroblasts extends beyond the promotion of fibrosis through  
361 myofibroblast induction. Macrophages act as potent sources of matrix metalloproteinases (MMPs), which can  
362 impact fibrosis directly via ECM degradation, or indirectly through cellular signaling and regulation of inflammation  
363<sup>137</sup>. The roles of MMPs in fibrotic control are extensive and highly context dependent. For example, MMP-9  
364 expression or activity has variably shown pro-fibrotic, neutral, or anti-fibrotic roles among different models of lung  
365 fibrosis<sup>137</sup>. The roles of MMPs in FBR-specific fibrosis are not entirely understood. Current understandings are  
366 summarized in **Table 2**. Additionally, macrophages can also limit fibrosis through IL-10 signaling, which reduces  
367 collagen production in activated myofibroblasts<sup>5,85,91,138–140</sup>. Arg1 metabolic flux in macrophages has also been  
368 associated with reduced fibrosis. Though the mechanism remains unclear, a possible explanation could involve  
369 'substrate stealing', wherein the highly metabolically active oxidative macrophages reduce available arginine for  
370 collagen production by fibroblasts<sup>45,84</sup>.

### 371 3. Advances in profiling and mitigating the foreign body response

372 Growing clinical evidence continues to establish causation and is demonstrating the significant burden of pathology  
373 attributable to the foreign body response. With increased attention, rapid advancements in both clinical and  
374 translational science are outlining new ways to interrogate and mitigate the FBR. Below, we outline exciting new  
375 approaches to profiling clinical disease, new *in vitro* and *in vivo* models of the FBR, and therapeutic strategies in  
376 development to avoid or treat FBR incidence.

#### 377 3.1. Investigating the foreign body response: models and techniques

378 As a complex phenomenon involving many lineages of cells and a highly context-dependent timeline, most  
379 mechanistic investigations of the FBR have benefitted from the use of animal models, informed by available clinical  
380 explants. These animal models and associated analyses have recently been expertly summarized in the context of  
381 hernia meshes, a well-studied and clinically significant example of the FBR<sup>141</sup>, with efforts to better model metabolic  
382 and biomechanical considerations often prompting the use of larger animal models<sup>142</sup>. With growing concern about  
383 the clinical adaptation of hernia mesh to pelvic organ prolapse, similar animal models of vaginal mesh application  
384 have been developed<sup>143,144</sup>, as has a model of neural implant FBR<sup>72</sup>, and models of silicone-implant associated  
385 capsular contracture<sup>29,145</sup>. Concerning the development of biocompatible materials to ameliorate the FBR, studies  
386 are often complicated by differences between and even within standardized animal lines by individual variability and  
387 by batch effects of implantation runs, both in terms of animal and implant batches<sup>57,146</sup>. The use of well-designed  
388 and statistically-informed experimental and surgical procedures can mitigate many of these complications, such as  
389 in the combinatorial screening of multiple materials in mice, which was successfully translated to primate usage<sup>147</sup>.  
390 From both model system and patient explants, single-cell techniques for functional profiling continue to be  
391 developed and standardized for replication and translation<sup>148</sup>; findings from multi-omics studies may be  
392 transferrable across species, experiments, and lines with the appropriate standardization and controls<sup>149</sup>.

393 From both *in vivo* models and with clinical explants, improved characterization techniques are helping to elucidate  
394 spatial relationships in the FBR. Cell profiling *in situ* with histology can allow for more precise profiling of cell  
395 phenotype and interactions in the local region<sup>43,55</sup>, while spatial transcriptomics of granulomas allow for cell



396 interaction profiling<sup>150</sup>. Intravital microscopy, by allowing repeated individual measures over time, allows for highly  
397 mechanistic studies to also inform factors of individualized response<sup>9</sup>. Individual genetics, medical history, and these  
398 intersecting and non-linear contributors to an individual's immune response are also more understood than ever,  
399 and inquiry and design with these concepts in mind will help to account for and harness individual variability in  
400 immunity<sup>151</sup>.

401 While *in vivo* models of the FBR provide valuable translational direction and holistic biometrics, they often fail to  
402 provide detailed mechanistic insight or quantitative metrics for optimization of therapeutic design. Immunology has  
403 benefitted significantly from recent advances in reproducible *in vitro* experimentation, and investigation of the FBR  
404 specifically is likely to benefit similarly from the level of mechanistic resolution that *in vitro* approaches provide. For  
405 example, new models of FBGC fusion *in vitro*<sup>42,152</sup> will allow for understanding of the factors leading to FBGC  
406 formation, as well as elucidate their functionality and amenability to productive control to then be trialled *in vivo*.  
407 Approaches to screening materials are well-established *in vitro*, although immunoregulatory applications are still  
408 very novel; macrophage M1/M2 differentiation was successfully profiled on combinatorial copolymer libraries in a  
409 recent study<sup>153</sup>, an approach that will prove instructive for further efforts. Similarly, statistically informed  
410 approaches reach a level of throughput that *in vivo* approaches cannot practically accommodate. Studies have  
411 revealed multifunctionality of individual cell phenotypes, and complex bidirectional relationships between  
412 macrophages and other stromal cells. Use of high-factor multivariate optimization encompassing genes, soluble  
413 excreted protein expression (e.g. VEGF), and microscopy-derived morphometrics (e.g. vascular tube formation and  
414 cell diameter) allowed for multiparametric optimization of macrophage-directed interstitial cell function *in vitro*<sup>154</sup>.  
415 These techniques can be used not only to model the peri-implant environment for material and therapeutic  
416 development, but also to derive quantitative multifactorial cause and effect mechanisms of immune function *in situ*.

417 As immunometabolism is a rapidly-growing field, so too are approaches by which to profile it. Traditional  
418 extracellular flux approaches such as those offered by Agilent Seahorse systems have successfully been used to  
419 profile specific pathway control in immune populations<sup>155,156</sup>. However, flow cytometry-based approaches to profile  
420 single-cell metabolism not only offer greater resolution of rare populations and limited samples *in vitro* or *in vivo*,  
421 but allow for correlation to cell phenotype by coupling to simultaneous traditional flow cytometric analyses, such as  
422 in the SCENITH method of profiling central carbon metabolism<sup>157</sup> and the QUAS-R method of profiling glutamine  
423 uptake<sup>158</sup>. Finally, metabolomic profiling allows for targeted pathway flux analysis<sup>159,160</sup>, which can be adapted for  
424 use in FBR characterization.

425

### 426 3.2. Material properties influencing the foreign body response

427 A degree of fibrosis is an expected outcome of wound healing, and thus will be present following implantation with  
428 any biomaterial. In fact, a physiological level of fibrosis is desired for mechanical stabilization and tissue integration  
429 of most implants<sup>161</sup>. Nevertheless, aberrant FBR outcomes are associated with costly implant revisions, highlighting  
430 the importance of understanding the dynamics of the biomaterial-host interactions. The extent to which material  
431 properties can alter fibrous encapsulation during the FBR has been outlined in previous reviews<sup>11,36,100,162</sup>. The  
432 mechanical properties of implanted materials are considered paramount to stable integration with host tissues.  
433 Mismatch between the stiffness of host tissue and an implant result in mechanical disruptions and worsen  
434 fibrosis<sup>12,163</sup>. Additionally, implant topography and scale can significantly alter the FBR. Jagged edges and  
435 macroscopic texturing that promote continuous tissue disruptions are associated with severe FBR outcomes<sup>11</sup>.  
436 Conversely, micro- and nano-texturing can impact adhesion and orientation of proteins, the patterning of cells, and  
437 ultimately mitigate adverse FBR outcomes<sup>164,165</sup>. A material's degradative profile is also believed to impact the FBR.  
438 Where material degradation is possible, implant fragmentation is conducive to both phagocytosis and discontinuous  
439 capsule formation, contributing to lower rates of capsular contracture and implant failure.



440 Material properties which influence the rate of protein adsorption at the implant interface are thought to be equally  
441 important in determining the outcome of the FBR. Some of the commonly identified characteristics that alter  
442 adsorption rates (and thus potentially affect the FBR) include surface roughness, surface charge and  
443 hydrophobia/hydrophilia<sup>5</sup>. Hydrophobicity/wettability will partially determine the degree of protein adsorption to  
444 its surface; hydrophilic surfaces resist protein adsorption relative to hydrophobic surfaces and are accordingly  
445 associated with less severe fibrotic reactions<sup>166,167</sup>. The surface charge of the material also in part determines the  
446 degree of protein adsorption to an implant, where increased charge increases protein adsorption<sup>168</sup>. Fibroblast  
447 traction based on both material stiffness and protein adsorption can influence macrophage migration and thus  
448 potentially chemotactic gradient formation, potentially creating positive feedback loops<sup>169</sup>; implants or factors that  
449 influence ECM deposition could therefore have outsized effects. Hence, implants of different materials might result  
450 in different levels of inflammation, and consequently the fibrosis following implantation. Understanding the  
451 metabolic differentiation in macrophages for different materials could elucidate some of the metabolic processes  
452 involved in a successful FBR outcome. However, the generation of particulates and degradation products from the  
453 material, or metabolites released from cells responding to the material, may result in local toxicity leading to  
454 unintended effects. Careful and systematic characterization of material disposition and mechanism of action of  
455 therapy will be required to ensure safety and efficacy of novel therapeutic materials.

456 Additionally, a better understanding of effector material properties will be critical in identifying potential targets for  
457 novel therapies, in addition to improving our understanding of the interplay between metabolism, signalling and  
458 effector function. Among the most commonly used medical implant materials are: poly(lactic-co-glycolic) acid  
459 (PLGA), polypropylene (PP), polyethylene (PE), polycaprolactone (PCL), polyethylene glycol diacrylate (PEGDA) and  
460 polydimethylsiloxane (PDMS). PP is preferentially used for permanent support-lending applications such as  
461 structural meshes and sutures<sup>170</sup>. PE implants are used in facial reconstruction as well as in joint replacement  
462 surgeries<sup>171</sup>. PCL implants are often used in cranioplasty<sup>172</sup>, and PDMS is commonly used for breast implants<sup>4</sup>. PLGA  
463 has a wide range of popular applications including tissue engineering scaffolding and micro- and nanoparticle  
464 delivery systems, is bio-degradable, and as such is considered to be highly biocompatible<sup>173</sup>. With a variety of  
465 materials available, each likely with inherent physicochemical properties, there is much to be learned about  
466 macrophages in the response to different IMD materials.

### 467 3.3. Biomaterial chemistry-based strategies to mitigate FBR 468 pathology: physical, chemical, and immune-signaling

469 Significant effort has been expended in profiling various physicochemical approaches to minimizing  
470 immunoreactivity and fibrosis to implanted materials. Alterations to physical properties outlined above, including  
471 wettability, stiffness, charge, (nano)topography and porosity have demonstrated clear trends in mitigating the FBR  
472 through altered rates of protein adsorption (**Figure 2**)<sup>162</sup>. Additionally, patterning of the surface that mimics features  
473 of the surrounding tissue has been shown to lead to a reduction of fibrosis through biomimicry, though this is not  
474 always possible or beneficial<sup>174</sup>. Changes to surface topography, such as micro-texturing, have demonstrated  
475 success in reducing implant-induced fibrosis *in vivo*, with relevant translation into the clinical setting for breast  
476 reconstructive surgeries. However, alarming links between micro-textured implants and the development of breast-  
477 implant-associated anaplastic large cell lymphoma (BIA-ALCL) have resulted in a market recall<sup>175,176</sup>. Similarly,  
478 polyurethane foam coatings for silicone breast implants have been associated with reductions in capsular  
479 contracture rates in cases where the coatings remain intact. Here, the porosity and texturing of the foam is thought  
480 to disrupt spatially continuous capsule formation and effectively dissipates mechanical tension<sup>177</sup>, albeit with risks  
481 of pathology such as ALCL as discussed above. However, concerns with regards to material degradation and toxicity  
482 have resulted in caution toward such implementations; the benefit of similar form designs with other materials  
483 remain undercapitalized.





484 Implant surface coatings have demonstrated the highest degree of success in reducing implant fibrosis. Zwitterionic  
485 materials/coatings and polyethylene glycol (PEG) surface conjugation have exhibited antifouling properties through  
486 decreasing adsorption of host proteins (i.e. ECM and DAMPs) to the implant surface<sup>35,114,178</sup>. This leads to a reduction  
487 in immune cell recruitment, macrophage recognition and activation, and proinflammatory cytokines<sup>147,179,180</sup>. Non-  
488 synthetic protein (e.g., gelatin and fibronectin)<sup>39,181,182</sup>, polysaccharide (hyaluronic acid, alginate, pectin, and  
489 heparin)<sup>147,182–187</sup>, and cytokine<sup>188</sup> polymer coatings have been used to reduce cellular activation by improving  
490 biocompatibility between the implant surface and tissue microenvironment. Similarly, use of a xeno/allo-derived  
491 adipose ECM extract showed improved biocompatibility and CD4 T-cell/M2 macrophage activation relative to gross  
492 fat when implanted in soft tissue<sup>189</sup>. While immunoevasion or biomimicry methods are useful in decreasing the  
493 acute FBR, they do not fully prevent host protein, ECM, and DAMP adsorption to the material, and downstream  
494 chronic immune cell activation. Therefore, strategies to target the FBR over long-term implantation are required;  
495 immunomodulation has become the major strategy to improve implant material outcomes in chronic settings<sup>190</sup>.

496 Research into immunomodulation to combat the FBR has been focused primarily on anti-fibrotic drug  
497 release/elution from material surfaces. In contrast to immunoevading materials, which either evade or integrate the  
498 inflammatory response, immunomodulating materials/systems actively suppress the inflammatory response through  
499 drug delivery. The simplest of these systems employ anti-fibrotic drug harbouring coatings which passively elute  
500 drug from the material surface<sup>191,192</sup>. Most of these models focus on the release of dexamethasone as a drug of  
501 choice due to its potent anti-inflammatory properties<sup>193–195</sup> and utility in the case of long-lasting implants<sup>196</sup>.  
502 Dexamethasone serves as an excellent model of coating chemistries and hydrophobic drug release<sup>13,146,193–196</sup>, and  
503 is highly effective at minimizing FBR. However, there are two key weaknesses in the translation of passive  
504 dexamethasone-eluting materials: 1) non-discriminating local immunosuppression is contraindicated for real-world  
505 surgeries due to the inevitability of occasional surgical site infections<sup>197</sup>, 2) dexamethasone impairs pro-regenerative  
506 responses<sup>72</sup>. Apart from dexamethasone, many other anti-fibrotic drugs such as methotrexate<sup>198</sup>, pirfenidone<sup>199</sup>,  
507 triamcinolone<sup>200–202</sup>, and tranilast<sup>203</sup> have shown efficacy in reducing implant induced fibrosis. Additionally, some  
508 non-traditional anti-fibrotic molecules such as kynurenic acid<sup>204</sup>, colony-stimulating factor 1-inhibitor GW2580<sup>205</sup>,  
509 NLRP3 inhibitor MCC950<sup>72</sup>, rapamycin<sup>206</sup> and sirolimus<sup>207</sup>, cytokines (e.g., IL-4 eluting materials)<sup>208</sup>, and  
510 immunotherapies (e.g., parasite antigens)<sup>209</sup> have all been shown to reduce implant induced fibrosis over various  
511 models.

512 Finally, justified concern of pathological fibrosis and surgical site infection has spurred the continued development  
513 of biologically derived implantable materials. These are most visible in applications for structural support, such as in  
514 surgical sutures and biological meshes for use in various hernia repairs (e.g., primary or incisional abdominal hernias,  
515 hiatal hernias, and inguinal hernias), as more specialized functional implants do not have biological options. Suture  
516 choice is heavily dependent on the tissue in question, as well as surgeon comfort or preference. Silk and gut sutures  
517 remain in active use for specialized indications; with the former being effectively permanent. Silk sutures offer  
518 advantages in handling, but are rarely strictly biologic in modern times as they often carry synthetic coatings. Overall,  
519 silk sutures offer little advantage in terms of both fibrosis (they remain highly immunogenic) and infection risk (as  
520 their braided design increases pathogen growth capacity)<sup>210–212</sup>. In terms of meshes for surgical plane repair,  
521 absorbable meshes, both biological and synthetic, have existed for decades. Biological meshes vary in terms of  
522 animal and tissue source, chemical crosslinking, and additives, and are designed with the intention to facilitate host  
523 tissue ingrowth before resorption renders them structurally noncontributory. These meshes are also often used in  
524 contaminated procedures in an effort to reduce the risk of surgical site infection, however infection rates,  
525 recurrence, and chronic pain suggest that biological meshes have no benefit over synthetics<sup>213–215</sup>. Absorbable  
526 synthetic meshes, although contentious in terms of their risk of hernia reoccurrence<sup>216</sup>, are generally growing in  
527 acceptance<sup>142,217–219</sup>, although judicious implant selection is still required when balancing patient demographics,  
528 anatomical considerations, and infection risks. In light of the high cost and uncertain benefit of biological materials  
529 in such a niche, it is likely that newer generations of chemically-defined synthetic products, both permanent and  
530 absorbable, will continue to grow in utility.



### 531 3.4 Immunometabolism as a promising therapeutic target to control the 532 FBR

533 Recent advancements in the field of immunometabolism highlight the potential of multiple small molecule  
534 metabolites to regulate the FBR and associated material-induced fibrosis <sup>220</sup>. Small molecules such as TCA  
535 metabolites have all shown promise in regulating the FBR <sup>159,221–225</sup>. In particular, the metabolite itaconate (IA) has  
536 emerged as a potent regulator of macrophage phenotype <sup>226–228</sup>. Itaconate has diverse direct and indirect  
537 immunomodulatory roles, including inhibition of both glycolytic and TCA pathways, multipronged inhibition of the  
538 NLRP3 inflammasome, and inhibition of myofibroblast activity, together allowing IA and its isomers to participate in  
539 highly specific contextual immunoregulation <sup>226,227,229–231</sup>. IA can target classically activated macrophages due to its  
540 potent anti-inflammatory effects, and can equally target alternatively activated macrophages by blocking monocyte  
541 differentiation into M(IL-4) <sup>222,232–234</sup>. In addition to TCA metabolites, glycolytic metabolites and analogues also  
542 demonstrate promising immunoregulatory activity. For example, incorporation of the glycolytic inhibitors 2-  
543 deoxyglucose and aminooxyacetic acid in subcutaneous polylactide (PLA) implants induced an anti-inflammatory  
544 phenotype <sup>235</sup>. PLGA has equally shown the ability to intrinsically exert metabolic control over local cells <sup>236</sup>. As  
545 mentioned above, when working with such chiral metabolites, validation of the stereochemistry of the formulation  
546 in question is crucial, and requires further standardization and control in studies <sup>237</sup>.

547 Despite its promise in precise and context-specific control of the immune response, immunometabolism presents  
548 several inherent challenges before it can be effectively leveraged in clinical practice. As demonstrated above,  
549 metabolites require relatively high concentrations before reaching therapeutic concentrations relative to many  
550 other bioactive molecules. This therefore poses a difficult engineering challenge to deliver doses safely to the target  
551 region. Chemical stability of bioactive components is also a consideration for all designs. Finally, these therapeutics  
552 need to be chemically and logistically compatible with the IMD in question. A potential solution to all of these  
553 individual challenges lies in the design of smart biomaterials. For example, metabolites such as IA require relatively  
554 high dose delivery to effectively modulate the immune system (5-10 mmol L<sup>-1</sup>) <sup>238,239</sup>. Passive drug-eluting surfaces  
555 fail to sustain effective high concentration drug delivery over long-term inflammation <sup>240,241</sup>. As such, with limited  
556 loading capacity, passive-drug eluting systems could be tailored to either deliver short-term high dose, or long-term  
557 low dose drug release. In the case of small molecule delivery, such as IA, a different approach to drug delivery needs  
558 to be taken to allow for high dose release in the long-term. IA, as well as other TCA metabolites, can be synthesized  
559 into the backbone of polyester polymeric biomaterials <sup>240</sup>. As the material is degraded in the host environment, the  
560 eluted metabolites regulate the fibrotic microenvironment. This method of drug delivery has two advantages over  
561 passive diffusion. Firstly, it greatly increases the loading capacity of the material <sup>241,242</sup>. Secondly, polymeric  
562 degradation can be tunable for temporal specific increased drug delivery <sup>242–244</sup>. Polymeric degradation kinetics rely  
563 on a variety of intrinsic polymer characteristics. Co-polymer ratio, molecular weights, polydispersity, material  
564 viscosity, transition temperatures, polymer endcaps, and hierarchical structuring (e.g., branching) are all useful tools  
565 in modulating degradation profiles and drug delivery <sup>245</sup>. This may provide a novel biomimetic avenue to reduce  
566 FBR-associated complications in future work.

## 567 4. Conclusions and future perspectives

568 Ultimately, the current understandings of FBR-associated fibrosis summarized in this paper indicate complex cellular  
569 heterogeneity and dynamic behaviours. The macrophage-fibroblast signalling axis is key to the development of  
570 capsular contracture and implant failures, but significant gaps remain in our understanding of the underlying  
571 mechanism and therefore potential targets. Critically, FBGC formation is a notable hallmark progression of the FBR  
572 but these cells share features of both M1 and M2 macrophages and defy canonical classification. Here,  
573 immunometabolism offers a promising new approach to cellular phenotyping in the implant microenvironment.  
574 Where the conventional immunological paradigm fails to capture the totality of cellular dynamics in the FBR,



575 evidence supporting links between metabolic behavior and pathological fibrotic signalling is abundant, underscoring  
576 the need for a comprehensive immunophenotyping of the tissue-implant microenvironment. Availability of effective  
577 treatment options to mitigate pathological fibrosis in the FBR is equally limited. The functionality of minor levels of  
578 fibrosis in improving tissue-implant integration, as well as the putative importance of early inflammation in  
579 promoting angiogenesis and tissue sterilization should caution the usage of broad-spectrum anti-inflammatory drugs  
580 such as dexamethasone. Additionally, currently available treatments lack tuneable release systems with sufficient  
581 reservoirs needed to promote extended release of anti-inflammatory drugs during the chronic stages of  
582 inflammation associated with pathological fibrosis in the FBR. The development of novel treatments is required to  
583 allow the temporal control of pharmacological interventions required to treat pathological inflammation and fibrosis  
584 in the FBR. Critically, FBR study is challenged by the lack of *in vitro* models available to recapitulate the complexity  
585 of the tissue-implant microenvironment, to overcome both the logistical and inherent immunological challenges in  
586 translating animal studies. Efforts are required to improve and develop better models of the FBR, with improved  
587 clinical translatability. Finally, major outstanding questions remain as to the clinical risk factors associated with  
588 undesirable FBR outcomes. Robust prospective analyses describing early patient phenotypes correlated to  
589 downstream clinical outcomes would help identify risk factors associated with FBR complications.

590 Ultimately, this review has served to summarize current understandings of the FBR to implanted biomaterials, as  
591 well as identify critical gaps requiring future investigation and development. This review has benefitted from  
592 significant recent attention to the mechanisms underlying immunometabolic regulation, as well as efforts in  
593 translating this knowledge to practical effect.

## 594 5. Data Availability Statement

595 No primary research results, software or code have been included and no new data were generated or analysed as  
596 part of this review.

## 597 6. Acknowledgements

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## 604 7. Author contributions

605 NIC conceived the manuscript. NIC, CR, ZF, KM wrote the manuscript. NIC, CR, ZF, KM and LDH edited the  
606 manuscript and approved the final version.

## 607 8. Conflict of interest

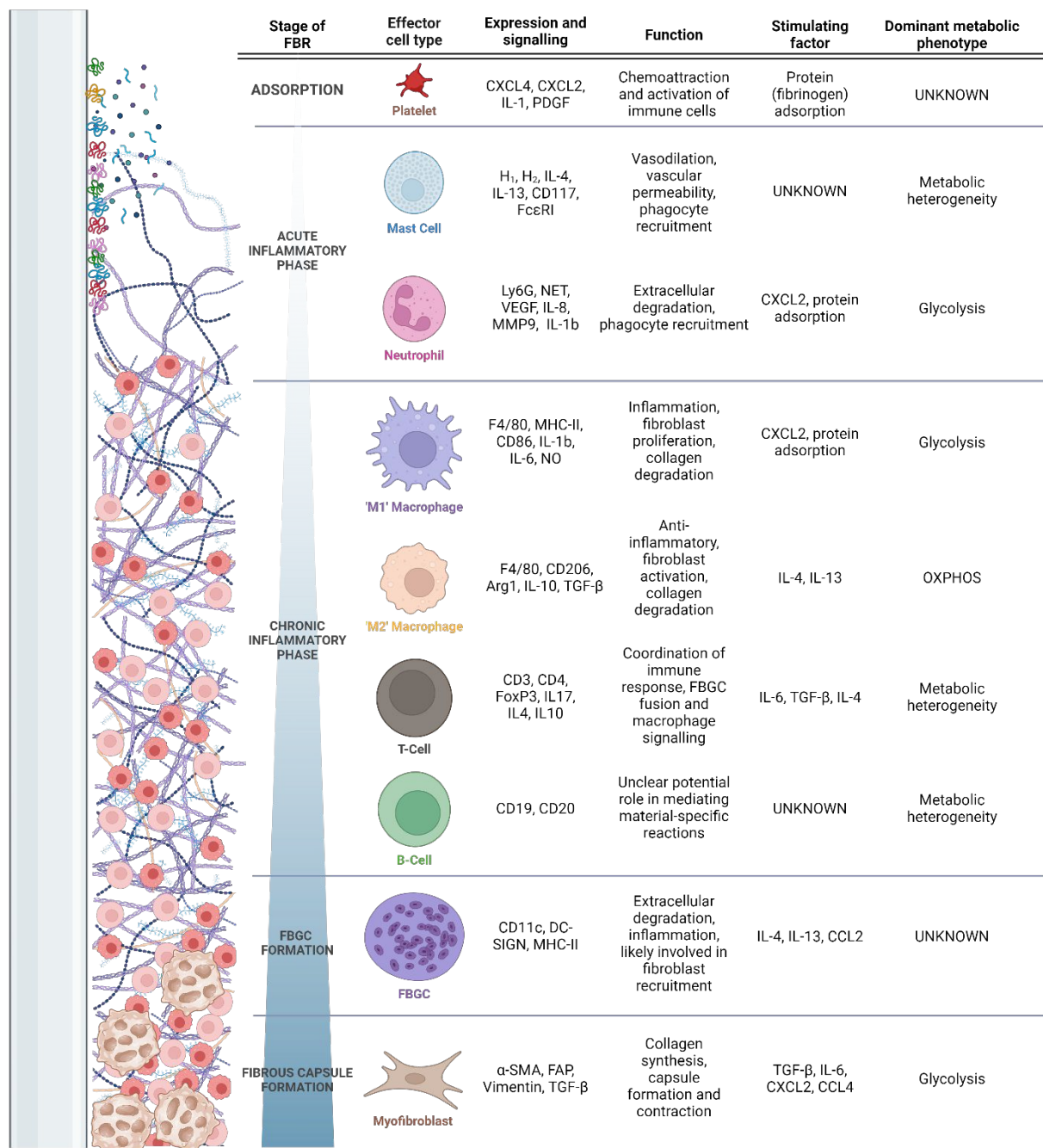
608 The authors declare no conflict of interest.

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610



## 611 9. Display items

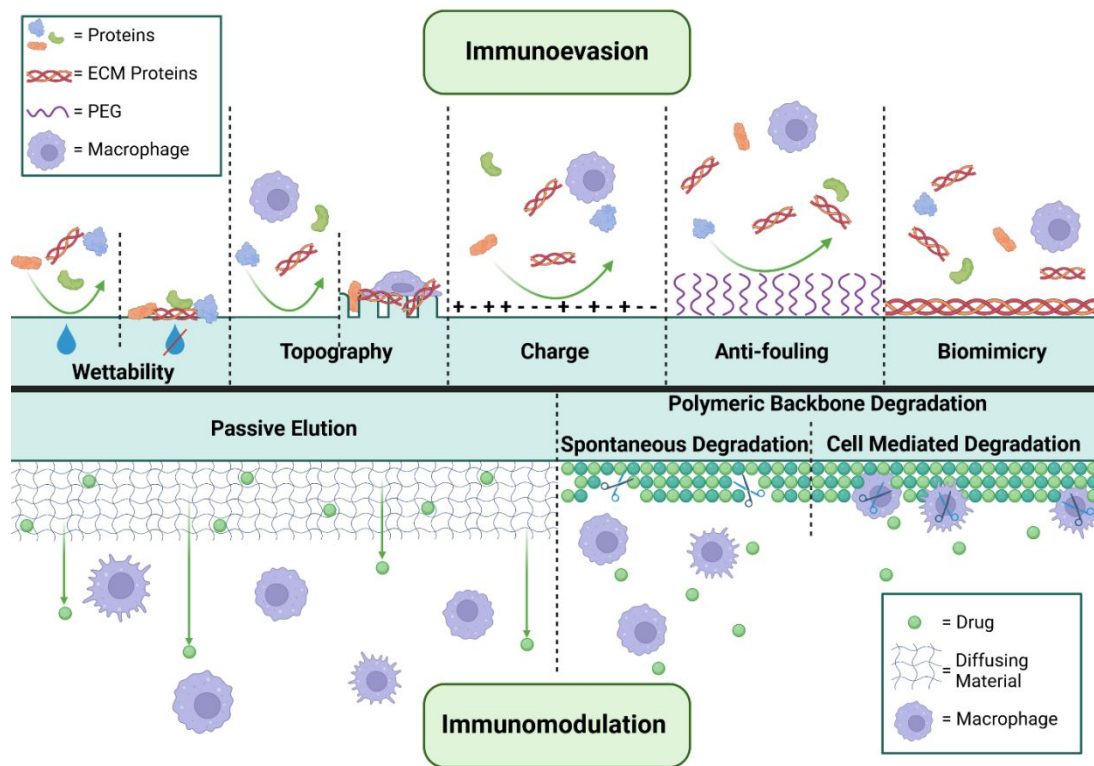


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613 **Figure 1:** Timeline of the FBR, recruitment of associated cells, and their secretory/metabolic phenotypes.

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**Figure 2:** Biomaterial design strategies to mitigate the immune-driven foreign body response.





618 **Table 1: Cytokines associated with the FBR.**

Cytokine	Cellular Origin	Target Cell	Function	Source(s)
TGF- $\beta$	Macrophages, fibroblasts	Fibroblasts, Macrophages	Promotes activation of fibroblasts. Promotes 'anti-inflammatory' profile in macrophages.	5,89,90
TNF $\alpha$	'M1-like' macrophages, neutrophils	Macrophages	Promotes differentiation and proliferation of myofibroblasts. Promotes inflammation.	129,130
PDGF	Macrophages, platelets, fibroblasts	Macrophages, Fibroblasts	Macrophage and fibroblast recruitment. Promotes myofibroblast differentiation and angiogenesis.	7,11,127,136
VEGF	'M1-like' macrophages, FBGCs	Endothelial cells	Promotes formation of neo-vasculature. Associated with elevated fibrosis.	9,11,136
IL-1 $\beta$	'M1-like' macrophages, neutrophils	Macrophages	Induces alterations to MMP secretion profiles. Promotes inflammation and expression of pro-fibrotic mediators.	131,132
IL-4	Mast cells, T-cells	Macrophages	Promotes alternative 'anti-inflammatory' activation of macrophages. Believed to play a role in cell fusion.	7,92,93
IL-6	Macrophages	Macrophages, Fibroblasts	Promotes activation of fibroblasts and stimulates paracrine TGF- $\beta$ signalling. Primes macrophages for anti-inflammatory signaling.	125,126,249
IL-10	'M2-like' macrophages	Macrophages, Myofibroblasts	Reduces collagen and $\alpha$ -SMA expression in activated fibroblasts. Promotes anti-inflammatory macrophage phenotype.	5,85,91,138–140
IL-13	Mast Cells, T-cells	Macrophages	Promotes alternative 'anti-inflammatory' activation of macrophages. Believed to play a role in cell fusion.	7,92,93
IL-17	Th17 cells	Neutrophils, Macrophages, Monocytes	Undifferentiated role in promoting chronic fibrosis. Associated with increased monocyte, macrophage, and neutrophil presence.	59



620 **Table 2:** Matrix metalloproteins (MMPs) associated with the FBR.

MMP	Classification	Macrophage phenotype	Co-associated genes/products	Temporal trend in expression	Niche/implant location	Species	Material	Technique	Reference
2	Gelatinase	(Gross analysis of explanted collagen disk)	IL-1 $\alpha$ , TNF $\alpha$ , TGF $\beta$	Progressive increase	Subcutaneous	Mouse	Bovine collagen	PCR gel	2
		CD163+/CD206+	Col-I	0.6-6 years post-implantation	Abdominal wall	Human	Polypropylene	Immunofluorescence microscopy	3
8	Collagenase	(Gross analysis of explanted collagen disk)	IL-1 $\beta$ , IL-10, CXCL1, CXCL2	Acute; decline after 1 week	Subcutaneous	Mouse	Bovine collagen	PCR gel	2
9	Gelatinase	(Gross analysis of explanted collagen disk)	IL-1 $\alpha$ , TNF $\alpha$ , TGF $\beta$	Appearance after 2 weeks	Subcutaneous	Mouse	Bovine collagen	PCR gel	2
		Proliferative macrophage ("MD2")	Proliferation-associated products	[Not characterized]	Subcutaneous	Rat	Silk	scRNAseq	4
		Adherent macrophage and FBGCs*	Adhesion-associated products	Significant increase in 3 days	<i>In vitro</i>	Human	PET, and modified PET	Antibody array and quantitative ELISA	246
		(Gross analysis of explanted capsule breast implant tissue)	Rac2	Increasing with increased Baker score	Breast	Human	Silicone	Bulk RNAseq	247
		FBGCs	Macrophage fusion, ECM degradation	[Not characterized]	Subcutaneous	Mouse	Mixed cellulose ester disks and polyvinyl alcohol sponges	Immunohistochemistry antibody array	248
12	Stromelysin	Giant cell ("C7")	ECM degradation, macrophage fusion, complement receptors, glycolysis	Progressive increase; resolution between 2-4 weeks in resorbable silk but not sponge	Peritoneal	Mouse	Silk and sponge (putatively cellulose)	scRNAseq	5
		Giant cell -like ("MD1")	ECM degradation	Intermediate	Subcutaneous	Rat	Silk	scRNAseq and histology	4
		Giant cell-like ("MD3")	Macrophage fusion, oxidative function	Progressive increase to 2 weeks	Subcutaneous	Rat	Silk	scRNAseq and histology	4

		Gross analysis of explanted capsule breast implant tissue)	Il-8, Tnsf11	[Not characterized]	Breast	Human	Silicone	qPCR	30
13	Collagenase	(Gross analysis of explanted collagen disk)	IL-1 $\alpha$ , TNF $\alpha$ , TGF $\beta$	Increasing after 2-3 weeks	Subcutaneous	Mouse	Bovine collagen	PCR gel	2
14	Membrane type	Epithelioid ("C6")	Pro-inflammatory (IL1b), ECM, chemokines	Progressive increase; resolution between 2-4 weeks in resorbable silk but not sponge	Peritoneal	Mouse	Silk and sponge (putatively cellulose)	scRNAseq	5
		(Gross analysis of explanted collagen disk)	IL-1 $\alpha$ , TNF $\alpha$ , TGF $\beta$	Increasing 2-3 weeks post implant	Subcutaneous	Mouse	Bovine collagen	PCR gel	2
		(Gross analysis of explanted CC breast implant tissue)	Rac2	Increasing with increased Baker	Breast	Human	Silicone	Bulk RNAseq	247
19	Not classified	Epithelioid ("C6")	Pro-inflammatory (IL1b), ECM, chemokines	Progressive increase; resolution between 2-4 weeks in resorbable silk but not sponge	Peritoneal	Mouse	Silk and sponge (putatively cellulose)	scRNAseq	5
MT3 (MMP 16)	Membrane type	Giant cell-like ("MD3")	Macrophage fusion, oxidative function	Progressive increase to 2 weeks	Subcutaneous	Rat	Silk	scRNAseq and histology	4
(Minimal)		M2-like (C5)	IL-10, chemokines	Acute, resolving in both resorbable suture and nonresorbable sponge	Peritoneal	Mouse	Silk and sponge (putatively cellulose)	scRNAseq	5



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## Data Availability Statement

No primary research results, software or code have been included and no new data were generated or analysed as part of this review.

