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# Poly Ethylene Glycol (PEG)-Related Controllable and Sustainable Antidiabetic Drug Delivery Systems

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## ABSTRACT

Diabetes mellitus is one of the most challenging threats to global public health. To improve the therapy efficacy of antidiabetic drugs, numerous drug delivery systems have been developed. Polyethylene glycol (PEG) is a polymeric family sharing the same skeleton but with different molecular weights which is considered as a promising material for drug delivery. In the delivery of antidiabetic drugs, PEG captures much attention in the designing and preparation of sustainable and controllable release systems due to its unique features including hydrophilicity, biocompatibility and biodegradability. Due to the unique architecture, PEG molecules are also able to shelter delivery systems to decrease their immunogenicity and avoid undesirable enzymolysis. PEG has been applied in plenty of delivery systems such as micelles, vesicles, nanoparticles and hydrogels. In this review, we summarized several commonly used PEG-contained antidiabetic drug delivery systems and emphasized the advantages of stimuli-responsive function in these sustainable and controllable formations.

**Keywords:** PEG, drug delivery system, antidiabetic, stimulating responsive release, sustainable release

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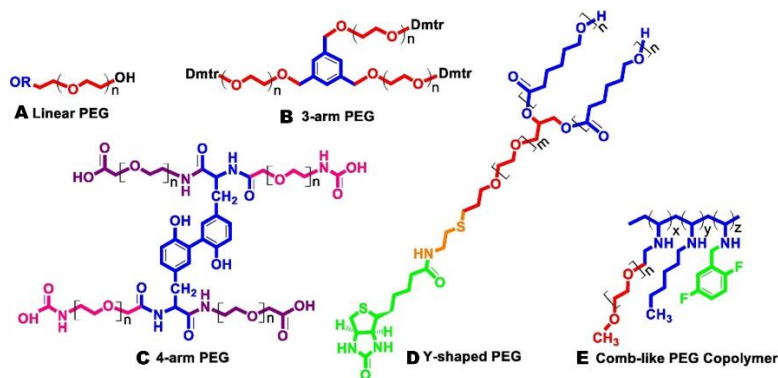
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## 1 Introduction

Diabetes mellitus is a severe chronic metabolic disease which imposes enormous burdens both personally and socially<sup>1</sup>. It is estimated that 4 million people died of diabetes and its complications in 2017, which has caused vast economic losses<sup>2</sup>. Generally, diabetes is classified into three subtypes: type 1 diabetes mellitus (T1DM) is an autoimmune disease, induced by the destruction of pancreatic  $\beta$ -cells which produce insulin<sup>3</sup>, type 2 diabetes mellitus (T2DM) is generated by the insufficient insulin supply or the insulin resistance<sup>4</sup>, gestational diabetes mellitus (GDM) is a glucose intolerance which first diagnosed during pregnancy<sup>5</sup>. Among all these subtypes, T2DM is the biggest threat to public health, the figure for T2DM patients almost accounting for 90% of the total<sup>6</sup>. A wide range of diabetes therapies including behavioral<sup>7</sup>, nutritional<sup>8</sup>, physical<sup>9</sup> and the most important, medicamentous therapies, have been adopted to alleviate diabetes and its complications.

Drug delivery systems (DDSs) plays a vital role in the clinical application of antidiabetic treatments since many antidiabetic drugs cannot achieve the ideal therapeutic effect without any assistance from a delivery system. Thus, appropriate DDSs are crucial in the designing of formulations. PEG-contained delivery systems are regarded as ideal options for the optimization of antidiabetic drug deliveries. PEG is a family of amphiphilic polymers<sup>10</sup> that sharing the same skeleton of repeating ethylene glycol units  $[(\text{CH}_2\text{CH}_2\text{O})_n]$  but with different molecular weights<sup>11</sup>. The most commonly used PEGylated reagents are the linear types of PEG chains. Except for the linear type, various shapes of PEG derivatives had been developed to meet different functions of specific attempts (**figure 1**). The derivation of PEG allowed more functionalization sites to be accessed, that broken through the limitations of the linear type PEG<sup>12</sup>, however, the derivation of PEG brought chemistry challenges in synthesis compared to the linear type PEG. Generally, PEGs have been extensively used in DDSs to ameliorate the physiochemical properties and bioactivity of various substrates<sup>13</sup>.



60  
61 **Figure 1.** PEG and multiple derivatives obtained via different linking methods such as linear (A), 3-arm (B), 4-arm (C),  
62 Y-shaped (D) and comb-like polymers (E)<sup>14</sup>.  
63

64 PEG molecules consisting of the DDSs are able to shield DDSs and drugs by their long hydrophilic  
65 chains. This distinctive architecture generally renders the delivery particles four characters: 1) enhanced  
66 enzymatic stability; 2) expanded hydrodynamic volume; 3) reduced immunogenicity of the large  
67 biomolecules and 4) decreased possibility of the macromolecules aggregation. These properties can  
68 decrease the kidney clearance, prolong the internal lifespan and promote the stability of particles, which  
69 lead to improved therapeutic effects<sup>15</sup>. Owing to its unique structure, PEG endows natural or artificial  
70 materials with improved physiochemical properties and biodegradability<sup>16</sup>.

71 Besides, due to the absence of inherent bioactivity, antidiabetic PEGylated polymeric systems can  
72 provide optimization of delivery and release process *in vivo* such as the prolonged half-life time<sup>17</sup>. Diverse  
73 types of PEGylated antidiabetic DDSs including nanoparticles, microspheres, nanovesicles, micelles and  
74 hydrogels are able to fit novel administration of drugs like transdermal<sup>18</sup> and oral administration<sup>19</sup> of  
75 insulin and intelligent insulin injections<sup>20</sup> which are able to overcome the disadvantages of traditional  
76 open-loop insulin delivery systems<sup>21</sup>. A number of studies have verified the advantages of PEGylated DDSs  
77 in diabetes treatments (**table 1**).  
78

**Table 1.** Part of the PEGylated DDSs applied in antidiabetic treatment

Loaded drug	DDS	Antidiabetic activity investigation model	Route of administration	Features
Insulin <sup>22</sup>	mPEG- <i>b</i> -P(GA- <i>co</i> -GPBA) micelles	Not applicable (N/A)	N/A	Glucose-responsive insulin release <i>in vitro</i>

Insulin <sup>23</sup>	PEG-PE micelles	BALB/c male mice	Intraperitoneal (i.p.) injection	Promote the renaturation of DTT-induced aggregated insulin
Insulin <sup>24</sup>	PEG/PNIPAM-P (Asp- <i>co</i> -AspPBA)/P (Asp- <i>co</i> -AspGA- <i>co</i> -AspNTA) complex micelles	STZ-induced mice	Subcutaneous (s.c.) injection	Protease resistance, improved blood glucose regulation capacity
Insulin <sup>25</sup>	PEG- <i>b</i> -P(AA- <i>co</i> -APBA)/P(AA- <i>co</i> -AGA) micelles	N/A	N/A	Enhanced glucose-responsive capacity
Insulin <sup>26</sup>	Microsphere loaded with CA-PLGA- <i>b</i> -(PEI-PEG) micelles-insulin conjugates	STZ-induced rats	S.c. injection	Reduced cytotoxicity, extended efficacy time
Insulin <sup>27</sup>	Insulin-deoxycholate composite micelles encapsulated by PCL-PEG-PCL vesicles	STZ-induced rats	S.c. injection	Improved encapsulation efficiency
Insulin <sup>28</sup>	PEG-P (Ser-ketal) nanovesicles	STZ-induced mice	S.c. injection	Acid-responsive insulin release <i>in vivo</i> with extended blood glucose regulation time
Insulin <sup>18</sup>	Microneedle array loaded with mPEG- <i>b</i> -P(SerPBE) vesicles	STZ-induced mice	Transdermal administration	Painless administration with glucose-responsive insulin release
Insulin <sup>29</sup>	Microneedle array loaded with PEG-poly (Ser-S-NI) vesicles	STZ-induced mice	Transdermal administration	Enhanced and stable blood glucose regulation ability with low hypoglycemia risk
Insulin <sup>30</sup>	DSPE-PEG-coated DPPC liposomes	STZ-induced rats	Femoral venous cannula	Improved incorporation efficiency and stability, more uniform particle size
Insulin <sup>31</sup>	PEG-PLGA nanoparticles	STZ-induced mice	S.c. injection	Sustain therapeutic effect with same minimum blood glucose level compared with free insulin
Insulin <sup>32</sup>	TPGS-emulsified PEG-capped-PLGA nanoparticles	STZ-induced rats	Oral administration	Oral administration validity of insulin, prolonged plasma glucose regulation time
Exenatide <sup>33</sup>	IgG Fc modified exenatide loaded by PEG-PLGA nanoparticles	Db/db mice	Oral administration	Obvious hypoglycemic effect compared with oral administration and SC injection of exenatide solution
Insulin <sup>34</sup>	Zn-insulin conjugates loaded by PLGA-PEG nanoparticles	N/A	N/A	Enhanced entrapment capacity with small particle size, sufficient stability for long-term storage
Insulin <sup>35</sup>	HP-55-coated capsules loaded with microparticles containing PLGA-lipid-PEG nanoparticles	STZ-induced rats	Oral administration	Oral administration validity of insulin, improved cell uptake, extended efficacy time with stable hypoglycemic effect
Plasmid DNA encoding GLP-1 <sup>36</sup>	Linear PEI/plasmid DNA nanoparticles coated by DPPC/1,2-dimyristoyl- <i>rac</i> -glycero-3-methoxy PEG-2000 (DMG-PEG)	Balb/c mice and db/db mice	Oral administration	Obvious blood glucose regulation induced by highly expressed GLP-1 gene transfected by NPs

Exenatide <sup>37</sup>	PEG- <i>b</i> -(PELG <sub>50</sub> - <i>g</i> -PLL <sub>3</sub> ) nanoparticles	STZ-induced rats	S.c. injection	Prolonged plasma duration time and hypoglycemic effect
Exenatide <sup>38</sup>	Zn-exenatide conjugates loaded by PEG-PLGA nanoparticles with low molecular weight protamine as oral absorption promoter	SD rats and db/db mice	Oral administration	Oral administration validity, extended efficacy time, higher plasma maximum concentration, enlarged AUC (3.27 folds), enhanced bioavailability compared with s.c. administration
Repaglinide <sup>39</sup>	PEG-PLGA nanoparticles	STZ-induced rats	Oral administration	Extended efficacy time
Insulin <sup>19</sup>	Calcium phosphate-PEG-insulin-casein particles	Female non-obese diabetic mice	Oral administration	Oral administration validity
Insulin <sup>40</sup>	Zn-insulin complexes loaded by PLA-PEG-PLA thermogel	STZ-induced SD rats	S.c. injection	Extended blood glucose regulation time, stable and prolonged plasma insulin concentration
Liraglutide <sup>41</sup>	PCGA-PEG-PCGA hydrogel	ICR mice and db/db mice	S.c. injection	Extended efficacy time
Lixisenatide <sup>42</sup>	PLGA-PEG-PLGA/PCGA-PEG-PCGA mixture hydrogel	ICR mice and SD rats	S.c. injection	Prolonged plasma lifespan, extended blood glucose regulation time, 7.6 folds larger AUC
Exenatide <sup>43</sup>	Depot-gel-in-microsphere-in-Matrix-gel system	SD rats	S.c. injection	Significant prolonged release time (46 days) with retained blood glucose regulation activity
Insulin <sup>44</sup>	4-arm-PEG acrylic hydrogel	N/A	N/A	Sensitive glucose-induced oxidation-degradation to achieve sustainable insulin release
Insulin <sup>45</sup>	Semi-IPN chitosan-PEG-pAAm hydrogel	N/A	N/A	Sensitive glucose-responsiveness, improved swelling ratio, drug loading capacity and entrapment efficiency derived from the increased PEG ratio
Insulin <sup>46</sup>	Arg-PEA/PEG-DA hybrid hydrogel with TD-1 as transdermal promoter	STZ-induced ICR mice	Transdermal administration	Good biocompatibility, moderate and prolonged blood glucose regulation capacity

79

80 As table 1 displayed, plenty of cases confirm the advantages of PEG as the ingredient of delivery  
81 systems. Among these fascinating systems, micelles, vesicles, nanoparticles (NPs), hydrogels and  
82 microneedles captured much attention due to their intrinsic properties. For example, micelles and vesicles

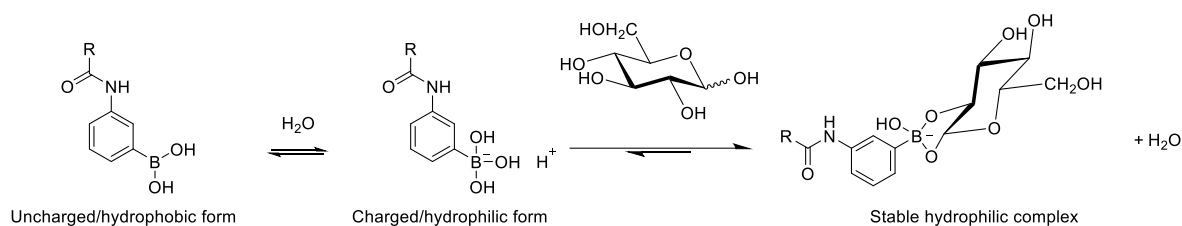
83 are able to encapsulate drugs, including but not limited to small-molecule drugs, peptides, proteins, DNA,  
84 with high efficiency to achieve efficient administration<sup>47</sup>. Microneedles array is considered as an effective  
85 transdermal administration of insulin since its painless, low cost and convenient for self-administration<sup>48</sup>.  
86 Pharmacokinetically, transdermal administration like microneedles matrix is able to bypass the "first-pass"  
87 elimination and the reduced maximum blood drug concentration also could minimize the risk of side  
88 effects<sup>49</sup>.

89 In recent decades, long-acting formulations with stable and controllable drug release profile are  
90 gaining much attention. These "intelligent" formulations generally require an adjustable release process to  
91 get with the physiological or pathological changes (like blood glucose concentration fluctuation,  
92 temperature variation or oxidative conversion of circumstance) to achieve the optimal dosage distribution  
93 in the whole release process. For instance, a severe side effect of conventional insulin injections is the  
94 excessive hypoglycemia induced by the burst release of insulin in the blood, the efficacy time is also  
95 restricted. By contrast, the release profile of intelligent DDSs is more moderate and persistent. Thus, the  
96 responsiveness to physiological or pathological changes, in the other words, the stimuli-responsive  
97 capacity is one of the crucial features for sustain release formulations.

98 Specifically, glucose-responsive capacity, which is able to inhibit the burst release of drugs to prevent  
99 undesirable hypoglycemia events and prolong the plasma glucose regulation time, is important for  
100 intelligent formulations in antidiabetic treatments. The glucose-responsive capacity of PEG-based DDSs  
101 can be achieved by introducing glucose-sensitive functions. For example, phenylboronic acid (PBA) is able  
102 to endow PEG-based DDSs with glucose-responsive capacity. As **scheme 1** demonstrated, there are two  
103 forms of PBA compounds in aqueous milieu<sup>50</sup>: uncharged/relatively hydrophobic form and  
104 charged/relatively hydrophilic form. Since charged borate is capable of covalently forming a stable  
105 hydrophilic complex with glucose through the esterification between boronic acid and *cis*-diol group. This  
106 reaction induces the hydrophilic conversion of PBA-contained components in aqueous milieu (like blood),



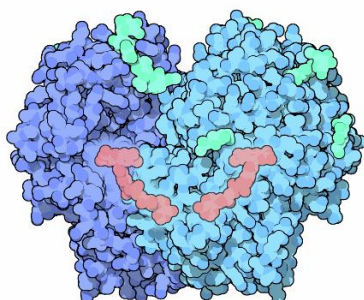
107 leading to the degradation of PBA-based micelles and the release of insulin loaded by micelles.



**Scheme 1.** The reversible formation of PBA-glucose complex

111 The earliest report of the reaction between glucose and PBA was reported by Kuivila, Henry G. *et al.*<sup>51</sup>.  
112 As time goes by, plenty of studies detailed and optimized the application of PBA in the antidiabetic DDSs.  
113 However, there are several challenges such as the discrepancy between apparent pKa for application and  
114 physiological pH<sup>52,53</sup> and the insufficient sensitivity to the fluctuation of blood glucose level<sup>54</sup> impede its  
115 further application. Therefore, a number of attempts have been deployed to form the accurate and  
116 adjustable glucose-responsive capacity to achieve controllable and sustainable drug release, providing  
117 more convenient formulations with better patient compliance.

118 Another case of stimuli-responsive capacity is glucose oxidase (GOx) which has been widely reported as  
119 a key component in the PEG-based DDSs. GOx is an enzyme that converts  $\beta$ -D-glucose and oxygen into  
120 gluconic acid and  $H_2O_2$ <sup>53</sup>. The intensive oxidation of  $H_2O_2$  can induce many reactions including the  
121 dissociation of oxidation-sensitive materials. Thus, a number of DDSs choose  $H_2O_2$  as the initiator of carrier  
122 degradation. However, the tissue inflammation induced by  $H_2O_2$  is a challenge of GOx-based  
123 glucose-responsive DDSs<sup>55</sup>. In general, the complexes of PEG and stimuli-responsive ingredients, because  
124 of their sustainable and controllable drug release capacity, are gradually being developed as a promising  
125 antidiabetic DDSs.



129 In this review, we discussed the application of PEG and PEGylated DDSs in the antidiabetic treatments  
130 by introducing several novel delivery systems and emphasizing the combination of PEGylated antidiabetic  
131 DDSs and stimuli-responsive capacity. In order to further illustrate the extensive applications of PEG in  
132 antidiabetic DDSs, several instances are provided below with a detailed description.

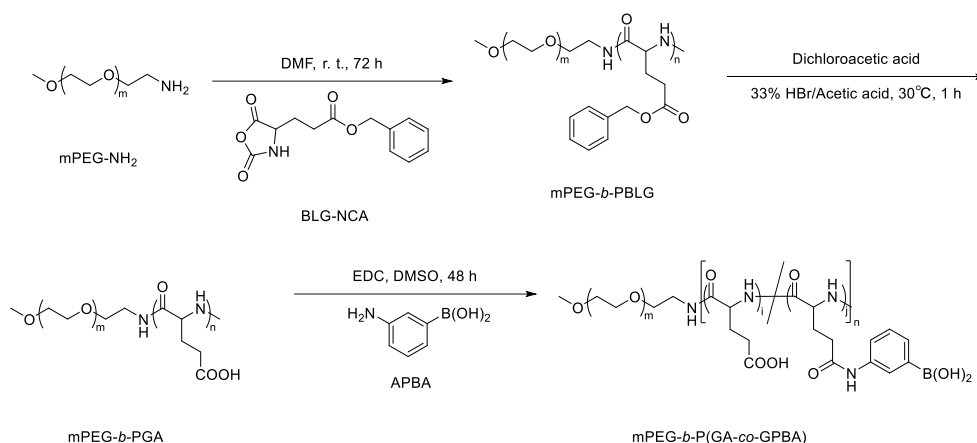
## 133 2 PEGylated micelles in antidiabetic treatments

134 Among all kinds of micelles, the amphiphilic block copolymeric micelle draws much attention as DDSs<sup>56</sup>.  
135 Amphiphilic block copolymeric micelles are a series of thermodynamically stable colloidal dispersions  
136 consisting of amphiphilic block copolymers, being of diverse strengths such as stabilizing drugs, targeting  
137 delivery, enhancing cellular uptake<sup>57,58</sup>. Three types of amphiphilic copolymeric micelles have been applied  
138 in the designing of DDSs: micelle-drug complexes are the composites of copolymeric micelles and drugs,  
139 micellar microcontainers trap drug molecules into their internal cavities to deliver them, and  
140 polyelectrolyte complexes are formed by the electronic interaction between cargoes and carriers, such as  
141 the conjugates of cationic block copolymers and polynucleotides<sup>59</sup>.

142 The amphiphilic block copolymeric micelles can be prepared by the self-assembling of the amphiphilic  
143 block copolymer chains which consist of two or more types of natural or synthetic polymers with different  
144 water affinity. The hydrophobic polymers consist of the internal side of the copolymeric chain and the  
145 hydrophilic polymers, like PEG, are placed on the other side to form external surface of micelles. In fact,  
146 PEG is considered as a popular ingredient to form amphiphilic micelles<sup>56</sup>. This unique architecture, widely  
147 known as the core-shell structure, allow micelles to deliver drugs with poor aqueous solubility in the water  
148 phase by loading them in the hydrophobic core of micelles<sup>60</sup>. In the PEG-contained amphiphilic block  
149 copolymeric micelles, the inner space of this spherical colloid encapsulates drugs and the outer PEG chains  
150 allow the micelle immune to unwanted results such as enzymolysis and aggregation<sup>61,62</sup>.

151 Many sorts of chemicals are reported in the development of PEG-contained amphiphilic block

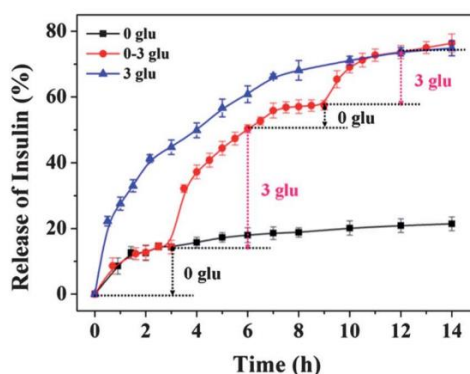
152 copolymeric micelles. For instance, synthetic polypeptides have been applied because they are highly  
 153 biocompatible and biodegradable. Li Zhao and coworkers chose poly (L-glutamic acid) (PGA) to prepare  
 154 monomethoxy PEG-*b*-poly (L-glutamic acid-*co*-N-3-L-glutamylamidophenylboronic acid) which  
 155 designated as mPEG-*b*-P (GA-*co*-GPBA) micelles by modifying mPEG-*b*-PGA with 3-amino phenylboronic  
 156 acid (APBA) to deliver insulin<sup>22</sup>. The whole synthetic route can be divided into two major steps (scheme 2).  
 157 Firstly, the copolymers mPEG-*b*-PGA were synthesized by the ring-opening polymerization (ROP) of  
 158  $\gamma$ -Benzyl-L-glutamate-N-carboxyanhydride (BLG-NCA) which followed by the debenzylation. Secondly,  
 159 APBA molecules were coupled with the pendent carboxyl groups of GA units to afford copolymers  
 160 mPEG-*b*-P (GA-*co*-GPBA). Afterwards, insulin was loaded into the hydrophobic core of the micelle via  
 161 dialysis method in deionized water.



162 **Scheme 2.** Synthetic route of copolymers prepared by Li Zhao *et al.*  
 163  
 164

165 The PBA groups in the polymers can interact with excessive blood glucose to form PBA-glucose  
 166 complexes. This hydrophilic variation allows the previously hydrophobic polymers to solve in the water  
 167 phase and the insulin loaded in advanced scatter in the local environment. Thus, these amphiphilic  
 168 copolymeric micelles synthesized in the work of Li Zhao *et al.* exhibited glucose-responsive and adjustable  
 169 drug release ability. As **figure 3** exhibited, When the insulin-loaded micelles were added to phosphate  
 170 buffer (PB) without any glucose (0 mg/mL), insulin released very slowly: only 12.6% of the total were  
 171 released within 3 h. After the concentration of glucose increased to 3.0 mg/mL, obvious release (37.7%)

172 was observed for the subsequent 3 h. Switching the concentration back to 0 mg/mL, insulin release was  
173 inhibited, only 7.2% amount was released in the following 3 h. Then, as the concentration return to 3.0  
174 mg/mL again, release behavior was recovered, verified by 16.1% release of insulin within 3 h.



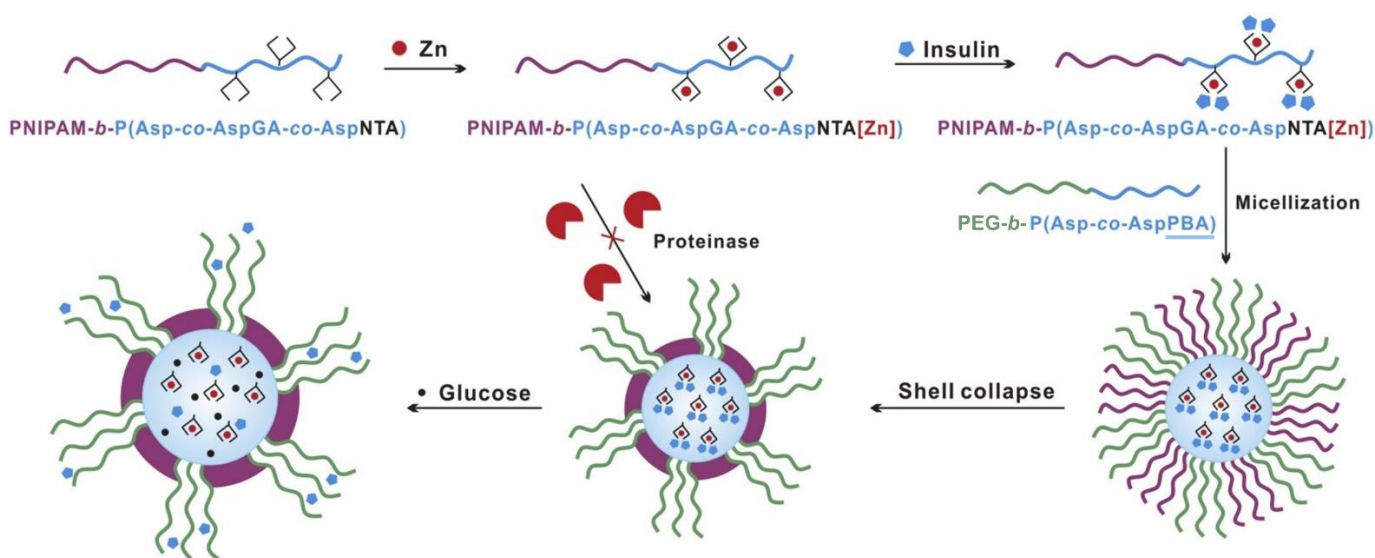
175  
176 **Figure 3.** The release profile of glucose-responsive mPEG-*b*-P (GA-*co*-GPBA) micelles

177  
178 In addition, no significant difference in circular dichroism (CD) spectra was observed between the  
179 released insulin and standard insulin sample, demonstrating that the preloaded insulin maintained its  
180 conformation after release. *In vitro* cytotoxicity investigation indicated the good biocompatibility of this  
181 copolymeric system.

182 This micelle system was prepared by Li Zhao *et al.*, all components of the micelle are biodegradable and  
183 biocompatible, allowing safely elimination after the release of insulin. Besides, except for functioning as a  
184 hydrophilic group, PEG is also able to balance the increased immunogenicity and protect micelles from  
185 proteases<sup>63</sup> in these copolymeric systems. CD spectra verified the therapeutic validity of insulin loaded  
186 inside micelles. All these features could support the promising prospect of these copolymeric micelles as  
187 antidiabetic DDS.

188 Although polypeptides are of many advantages, a fatal demerit of these materials is the vulnerability to  
189 proteases which may weaken the protective effect of micelles on the loaded drug. Thus, Wu Gang and  
190 coworkers developed complex poly amino acids micelles with proteases resistance<sup>24</sup>. As **figure 4** showed,  
191 PEG and poly (N-isopropylacrylamide) (PNIPAM) complex were employed as the composite shell, while  
192 poly (aspartic acid-*co*-aspartic acid phenylboronic acid) designated as P(Asp-*co*-AspPBA) and poly

193 (aspartic acid-*co*-aspartic acid glucosamine-*co*-aspartic acid nitrilotriacetic acid) designated as  
 194 P(Asp-*co*-AspGA-*co*-AspNTA) functioned as the composite core. Insulin was connected to the divalent zinc  
 195 ions which coordinated with NTA groups located on the P(Asp-*co*-AspGA-*co*-AspNTA) copolymeric chains.  
 196 The glucose sensitivity of these composite micelles was derived from the PBA/GA complexation. PNIPAM  
 197 played a crucial role in the protection of micelles from proteolysis by collapsing to form a hydrophobic  
 198 shield. As expected, the resistance to proteases and improved blood glucose regulation capacity was  
 199 observed. Under abnormally high blood glucose level, the interaction between glucose and PBA moieties  
 200 gradually leads to the swelling and disassembly of complex micelles and the release of insulin.



201  
 202 **Figure 4.** Schematic of the preparation and release process of complex micelles prepared by Wu Gang *et al.*<sup>24</sup>  
 203

204 Rujiang Ma *et al.* devised a type of glucose-responsive complex micelle consisting of block copolymer  
 205 PEG-*b*-poly (acrylic acid-*co*-acrylamidophenylboronic acid) (PEG-*b*-P(AA-*co*-APBA)) and glycopolymer  
 206 poly (acrylic acid-*co*-acrylglucosamine) (P(AA-*co*-AGA))<sup>25</sup>. In this complex micelle system prepared by  
 207 Rujiang Ma and coworkers, PEG chains on the outer layer functioned as a hydrophilic shell against  
 208 aggregation. Another ingenious strategy is the introduction of glycopolymer: the most suitable pH for  
 209 glucose-responsiveness is the apparent pKa of PBA (around pH 9) which is much higher than physiological  
 210 pH (around 7.4), so the glucose sensitivity is restricted under physiological condition. Due to the  
 211 complexation between PBA and glycopolymer, the apparent pKa of PBA was decreased and the glucose

212 sensitivity was enhanced.

213 Except for the enzyme degradation, undesirable aggregation is another obstacle to insulin delivery and  
214 application. For example, insulin amyloid deposition has been found at the sites of frequent subcutaneous  
215 insulin injections<sup>64</sup>. This objectionable aggregation generally means the weakened hypoglycemic activity  
216 of insulin<sup>23</sup> and the increase of its immunogenicity<sup>65</sup>. Besides, these deposits also cause trouble in the  
217 production, storage and transportation of insulin<sup>66</sup>.

218 In recent decades, nanocage is gradually concerned as an alternative to traditional DDSs and  
219 PEGylation is regarded as an effective method of nanocage functionalization to obtain various advanced  
220 properties<sup>67</sup>. A PEG-phosphatidylethanolamine (PEG-PE) micelle system was developed to achieve the  
221 reversion of insulin aggregation, inspired by the GroEL-GroES chaperonin system of *Escherichia coli*<sup>23</sup>.  
222 These diblock copolymeric micelles functioned as nanocages to concentration-dependently reverse the  
223 dithiothreitol (DTT)-induced insulin aggregation.

224 Insulin is a hypoglycemic protein with 51 amino acids, composed of two chains designated as A chain  
225 with 21 amino acids and B chain with 30 amino acids. These two chains are connected by two disulfide  
226 bonds between A and B chains. DTT can cut off these two disulfide bonds to afford separated unfolded  
227 peptide chains and the interaction between the hydrophobic parts of these chains finally produces the  
228 aggregations. Specifically, these nanocages were able to trap A and B chains of insulin cleaved by DTT,  
229 screening the interaction between their hydrophobic moieties which is the main factor of sedimentary  
230 formation. Besides, the separated A and B chains were able to reconnect with each other to afford native  
231 insulin with hypoglycemic activity. This process can be verified by the MALDI-TOF mass spectra and  
232 hypoglycemic effect in mice. CD spectroscopy suggested that PEG-PE micelles stabilized the secondary  
233 structure of native insulin, preventing chains from false folding.

234 Jun Wang and coworkers devised and prepared a type of cholic acid (CA)-PLGA-*b*-(polyethyleneimine  
235 (PEI)-PEG) micelles to load insulin on their surface through electrostatic interaction (**figure 5**)<sup>26</sup>. An

236 attractive point of this work reported by Jun Wang *et al.* is the combination of PLGA, PEI and PEG. PLGA has  
237 been considered as an ideal material to build micro/nano structure for drug delivery<sup>68,69</sup>. However,  
238 according to existing reports, the degradation of PLGA could produce an local acidic atmosphere<sup>70</sup> which  
239 may cause negative effects on the loading proteins and peptides but can be ameliorated by introducing  
240 PEI into the delivery system<sup>71</sup>. However, the high cytotoxicity of PEI with large molecular weight (such as  
241 25kDa) derived from its excessive positive charge impedes its application<sup>72,73</sup>. Thus, PEG was introduced  
242 into the system combining with low molecular weight PEI to ensure the safety of polymeric delivery system,  
243 and the validity of these strategies have been confirmed<sup>73,74</sup>. Insulin was able to efficiently loaded on the  
244 cationic polymeric micelles via the electrostatic force between the abundant cations on PEI-PEG layer and  
245 insulin. Extended blood glucose regulation time was observed in *in vivo* investigation.

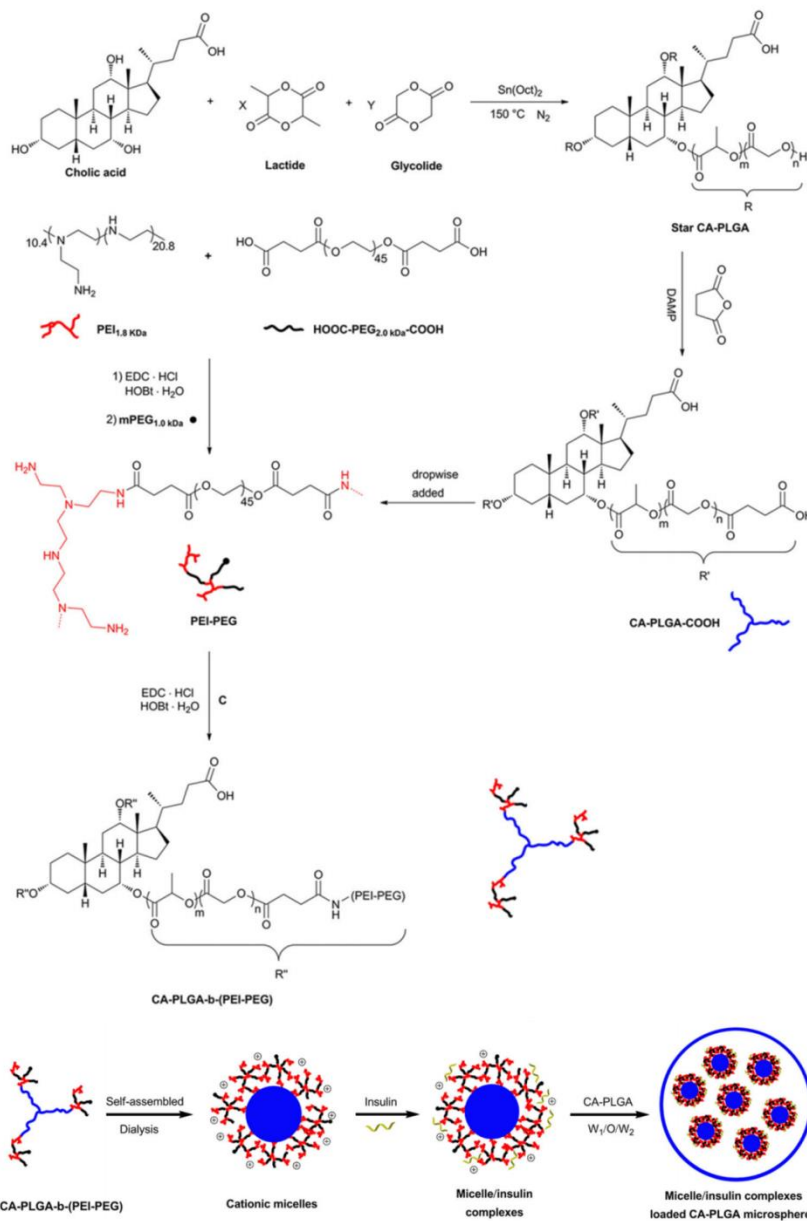


Figure 5. Preparation of the microsphere loaded with CA-PLGA-*b*-(PEI-PEG) micelles-insulin conjugates<sup>26</sup>.

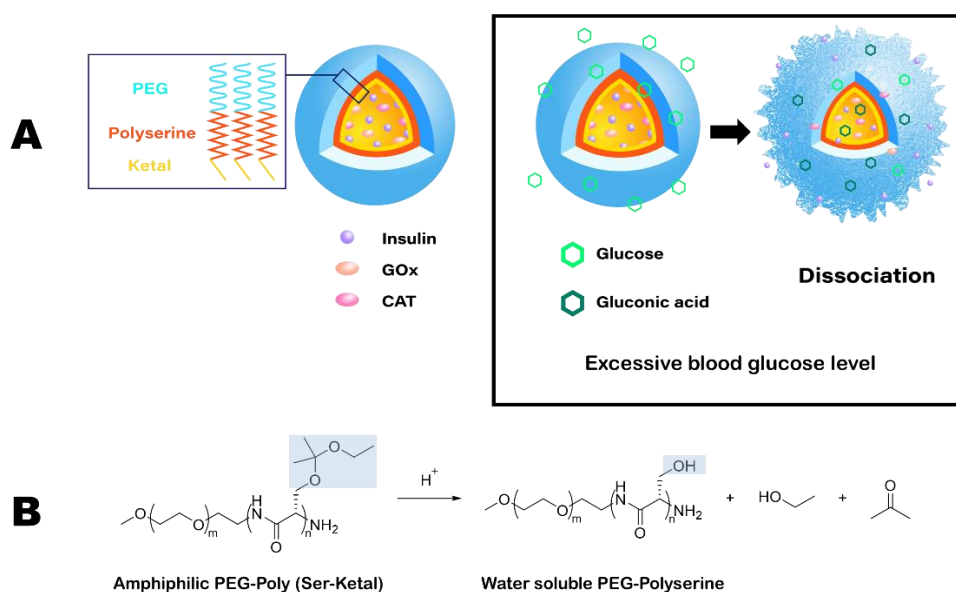
### 3 PEGylated vesicles in antidiabetic treatments

Vesicles are a series of particles sharing a similar structure which consist of a lipid bilayer membrane and an internal hollow space separated from the outside. The bilayer membrane is composed of the hydrophilic "heads" which are generally forming the surface of the membrane and hydrophobic "tails" which are buried under "heads". Biologically, vesicles including liposomes and exosomes are of great importance in the transportation, communication and other metabolic processes of various cells. The potential of vesicles as drug delivery systems has been widely reported<sup>75-77</sup> such as paclitaxel-loaded exosomes modified by PEG and ligand treating non-small-cell lung carcinoma (NSCLC) with prolonged



257 circulation time<sup>78</sup>.

258 Inspired by bio-generated vesicles, synthetic ones are also have been developed to deliver drugs. For  
259 instance, inspired by native vesicles, Wanyi Tai *et al.* devised a biomimetic polymersome nanovesicle  
260 system with an acid-sensitive capacity<sup>28</sup>. This copolymeric vesicle employed PEG as the hydrophilic "heads"  
261 and ketal-modified polyserine (PEG-P (Ser-Ketal)) as the hydrophobic "tails" to form the bilayer membrane.  
262 Insulin, glucose oxidase (GOx) and catalase (CAT) were loaded in the hollow space of the vesicle. Drugs  
263 were well encapsulated while glucose molecules, due to their small size, were able to penetrate into the  
264 inside of the copolymer membrane. Afterwards, the interaction between glucose and GOx afforded  
265 gluconic acid and H<sub>2</sub>O<sub>2</sub>, leading to the local pH decrease. Meanwhile, H<sub>2</sub>O<sub>2</sub> generated from the  
266 aforementioned process was converted to oxygen by CAT to avoid damaging other cellular components  
267 and the deactivation of GOx<sup>79-82</sup>. As the result of pH decrease, the ketals located on the polyserine  
268 segments of copolymers dissociated through acidic hydrolysis, causing hydrophilic conversion (**figure 6 B**)  
269 of entire copolymers in aqueous phase (like blood). Water-soluble copolymers without ketal moieties  
270 solved in the solution and the copolymeric membrane gradually fractured. Finally, insulin loaded in  
271 advance was released (**figure 6 A**).



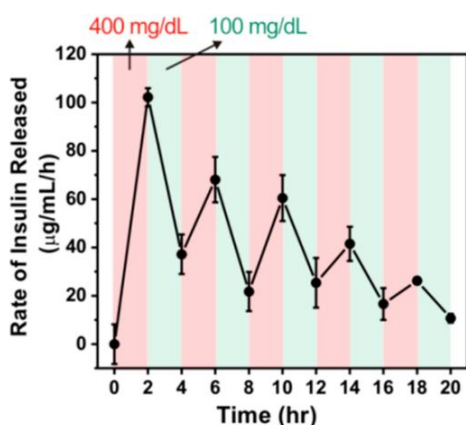
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273

274

**Figure 6.** Schematic of the degradation of acid-sensitive diblock copolymer nanovesicles.

275 The PEG-contained copolymeric nanovesicles were prepared by Wanyi Tai *et al.* exhibiting the  
276 glucose-responsive capacity. A low level of insulin release was observed under 100 mg/dL glucose or  
277 glucose-free condition in PBS buffer over 12 h, compared with the rapid release under hyperglycemic  
278 condition. More importantly, with alternative conversion between normoglycemia and hyperglycemia  
279 every 2 h, insulin release exhibited a pulsatile trend correspondingly (figure 7).



280  
281 **Figure 7.** Pulsatile insulin release profile of PEG-P (Ser-Ketal) nanovesicles.  
282

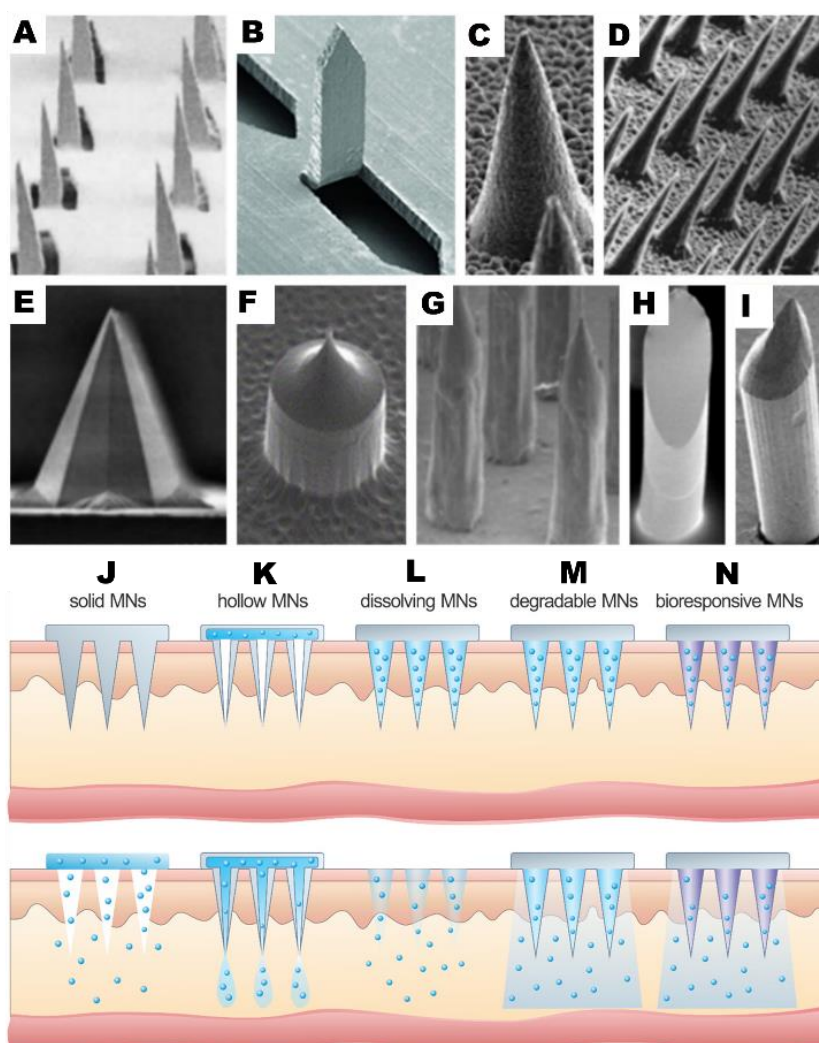
283 Difference up to 3-fold in release rate was observed when the glucose concentration altered. However,  
284 this trend gradually diminished as this circulation continued. A possible reason could be the gradual  
285 dissociation of vesicles and the leakage of enzymes.

286 Except for Wanyi Tai and coworkers, the applications of vesicles in the antidiabetic DDSs are also  
287 studied by others. Anna Kim *et al.* reported a distearoylphosphoethanolamine-PEG (DSPE-PEG)-coated  
288 1,2-dipalmitoyl-sn-glycero-3-phosphocholine (DPPC) liposomes to reduce the uptake of liposome by  
289 reticuloendothelial system (RES) in parenteral administration<sup>30</sup>. After coated by DSPE-PEG, the liposomes  
290 were of more uniform size with enhanced aggregation resistance, and the circulation time was also  
291 extended by 1 h.

292 In addition to individually applied, vesicles are also able to combine with other delivery systems.  
293 Rachna Rastogi *et al.* developed a poly (caprolactone)-PEG-poly (caprolactone), which designated as  
294 PCL-PEG-PCL, copolymeric vesicle system to encapsulated insulin-deoxycholate composite micelles<sup>27</sup>.

295 Compared with free insulin, the encapsulation efficiency of complex micelles was enhanced by around  
296 10-50%. Burst insulin release was weakened and the efficacy time was prolonged by 2 h, but the increased  
297 hydrophobicity of the delivery system exerted a negative influence on its pharmacological effects which  
298 emphasize the critical role of equilibrium between hydrophilicity and lipophilicity.

299 Another case is microneedles-vesicles composite delivery system. Microneedles (MNs) have been  
300 introduced to antidiabetic DDSs to achieve painless transdermal administration which is able to promote  
301 patient compliance due to their unique properties<sup>53</sup>. MNs (**figure 8**) have been extensively explored for the  
302 transdermal administration of various substances such as small molecule drugs<sup>83</sup>, proteins<sup>84</sup> and particles<sup>85</sup>,  
303 and diverse type of MNs including hollow, solid, coated, dissolving and hydrogel forming<sup>49</sup> are also  
304 developed by various materials. The application of MNs in antidiabetic DDSs have been systematically  
305 reported<sup>48,49</sup>.

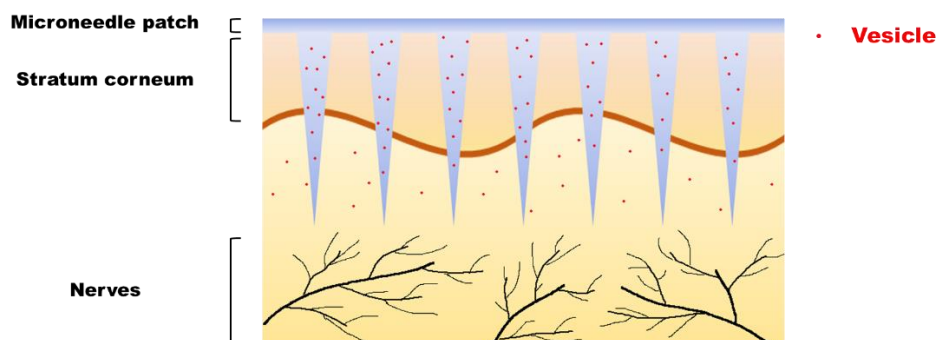


306  
307

**Figure 8.** Images of different shapes of MNs (A-I)<sup>48</sup> and different type of MNs including solid MNs (J), hollow MNs (K),

dissolving MNs (L), degradable MNs (M), and bioresponsive MNs (N)<sup>86</sup>.

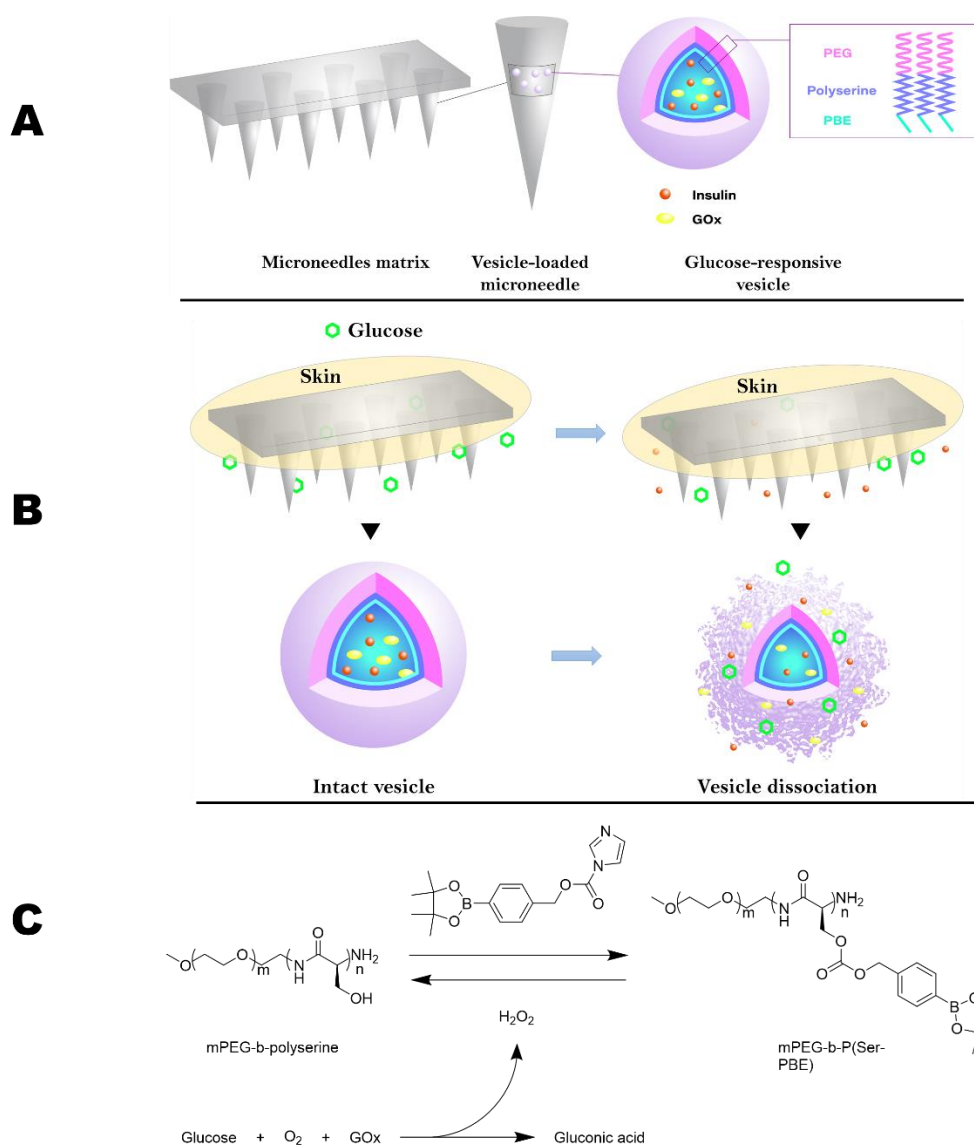
The painless and efficient transdermal delivery capacity of MNs is derived from their unique architecture. As **figure 9** displayed, stratum corneum (SC) with a thickness of 10–15  $\mu\text{m}$  is considered as the main obstacle for transdermal administration. According to reports, only sufficiently lipophilic substances with molecular weight lower than 500 Da<sup>87</sup>, which is too small for most of the vesicles, could diffuse into SC. As the result, common transdermal administration is restricted for vesicles. Fortunately, nerves are located a few hundred microns below SC<sup>88</sup>. Thus, the painless administration can be achieved through bypassing the SC without or slightly touching nerves, inducing little or no pain. Besides, compared with conventional transdermal formulations which are highly restricted by the diffusion limit of SC, MNs are able to efficiently deliver the diver types of drugs without SC impedance. The introduction of MNs allow the vesicles which are too large to penetrate SC by self-diffuse to cross this screen with high efficiency, and glucose-responsive vesicles endow formulations with sustainable and controllable release, achieving optimal dosage distribution. The MNs-vesicles complex delivery systems integrate both the efficient transdermal administration and glucose-responsive capacity.



**Figure 9.** The transdermal administration of vesicles performed by microneedles array.

Xiuli Hu *et al.* integrated self-assembling amphiphilic block copolymeric vesicles which were composed of PEG, polyserine and phenylboronic ester (PBE) with microneedles array to form a MNs/vesicles complex delivery system to achieve sustainable and controllable release<sup>18</sup>. Cross-linked hyaluronic acid (HA) was adopted to form the microneedle structure. **Figure 10 A** shows the architecture of mPEG-*b*-P (Ser-PBE)

330 vesicles-loaded MNs array. **Figure 10 B** exhibits that excessive blood glucose concentration causes the  
 331 disassociation of vesicles and the consequent insulin release. **Figure 10 C** displays that the origin of  $H_2O_2$   
 332 which was directly responsible for the dissociation of vesicles is the interaction between GOx encapsulated  
 333 in the vesicles and the glucose penetrating into the inside of vesicles.



334  
 335 **Figure 10. A** exhibits the architecture of MNs loaded with mPEG-b-P(Ser-PBE) vesicles; **B** displays the glucose-induced  
 336 insulin release; **C** is the degradation reaction of copolymeric vesicles.

337  
 338 In the presence of  $H_2O_2$ , copolymers lost their PBE groups and transformed to hydrophilic molecules,  
 339 being able to solve in the aqueous phase. Consequently, gradual degradation of vesicles led to the release  
 340 of preloaded insulin. Xiuli Hu *et al.* adopted the strategy mentioned above to endow the vesicles with  
 341 glucose-responsive function: GOx was introduced to catalyze the conversion from glucose to gluconic acid

342 and afford  $H_2O_2$  which is responsible for the disassembly of vesicles. These vesicles were encapsulated into  
343 microneedles formed by cross-linked HA, since HA is highly biocompatible with appropriate stiffness to  
344 penetrate skin. Once the plasma glucose level was abnormally high, surplus glucose penetrated the HA  
345 membrane into the inside of vesicles, reacting with GOx to yield  $H_2O_2$ , leading to the dissociation of  
346 vesicles and the release of insulin loaded in advance. The *in vivo* influence was carefully evaluated: the  
347 biocompatibility of this formulation was acceptable, along with the negligible hypoglycemic risk. Rapid  
348 glucose-responsive insulin release was observed under hyperglycemia and once reaching normoglycemia,  
349 the release rate was restrained. On the one hand, the glucose-sensitive drug release capacity of this MNs  
350 loaded with mPEG-b-P(Ser-PBE) vesicles reduce the risk of undesirable side effects such as hypoglycemia  
351 and potential damage induced by  $H_2O_2$  to cells; on the other hand, this MNs allows the optimized dosage  
352 distribution for drug release, significantly prolonged blood glucose regulation time and administration  
353 interval.

354 In another case of the composite MNs delivery system reported by Jicheng Yu and co-workers<sup>29</sup>, a type  
355 of MNs matrix containing hypoxia and  $H_2O_2$  dual-sensitive vesicles was developed based on a similar  
356 strategy. The loaded copolymeric vesicles, designated as PEG-P (Ser-S-NI) vesicles, were composed of PEG  
357 and polyserine modified by 2-nitroimidazole through thioether bridge. The  $H_2O_2$  and hypoxia-responsive  
358 capacity derived from the thioether and 2-nitroimidazole respectively. The thioether was transformed to  
359 sulfone after reacting with  $H_2O_2$ , this oxidation leading to the hydrophilic conversion of polymer<sup>89,90</sup>.  
360 Simultaneously, 2-nitroimidazole converted to hydrophilic 2-aminoimidazole under hypoxia condition  
361 which was mediated by the transformation from glucose to gluconic acid<sup>91,92</sup>. In this composite system, the  
362 hypoxia-responsive 2-nitroimidazole parts enhance the glucose sensitivity of delivery systems, and the  
363  $H_2O_2$ -responsive thioether parts consume excessive  $H_2O_2$  generated by GOx, avoiding negative effects  
364 including inflammation.

365

#### 4 PEGylated nanoparticles (NPs) in antidiabetic treatments

The benefits of PEGylation to NPs have been systematically summarized<sup>93</sup> and verified by many studies<sup>20,31-34,36,38,39,94-96</sup>. Basically, PEG as ingredients could resolve the instability of insulin in the harsh formulation conditions, while poly (D, L-lactide-co-glycolide acid) (PLGA) is a type of polymer that has already been successfully applied in biomacromolecule delivery<sup>69</sup>. Yusuf Haggag and coworkers employed double emulsion technique to prepare a series of NPs consisting of poly (D, L-lactic-co-glycolic acid) (PLAG)-PEG diblock polymers to optimize the entrapment efficiency of insulin<sup>31</sup>. According to the experimental results, part of PLGA-PEG polymers via homogenized insulin-loading method optimized their entrapment efficiency and release kinetics of insulin. The *in vivo* studies revealed the aggregation resistance, improved stability and the maximum retention of hypoglycemic bioactivity of insulin during the whole release process. Moreover, the PLGA-PEG diblock copolymeric NP system devised by Yusuf Haggag *et al.* exhibited a marked insulin sustainable release manner.

A similar PLGA-PEG NPs system was designed by Sunandini Chopra and coworkers<sup>34</sup>. The difference was that they added zinc ions to insulin to form the insulin-Zn hexamers via the chelation first, then co-assembled with PLGA-PEG under a carefully adjusted pH nanoprecipitation condition to form NPs. Finally, after a PBS washing process, the insulin-Zn PLGA-PEG NPs were afforded (figure 11). NPs prepared via this complexation between zinc ion and insulin exhibit significant improvement (about 10-fold) in insulin loading capacity.

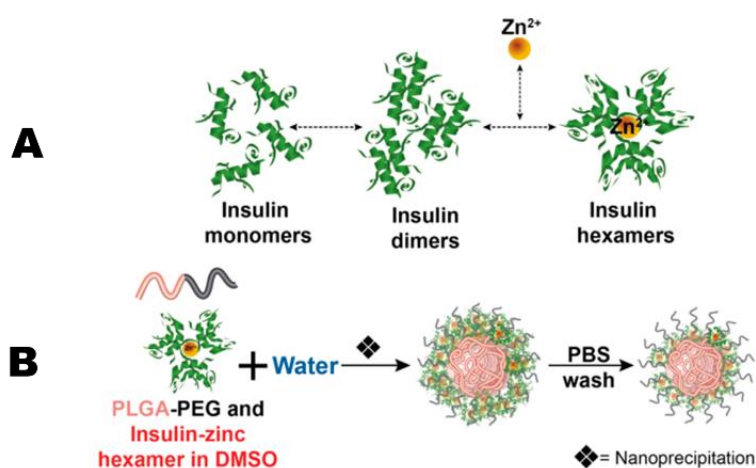


Figure 11. Schematic of the preparation of insulin-Zn complex and the NPs developed by Sunandini Chopra and

In the work of Shelesh JAIN and Swarnlata SARAF, PEG significantly extended the efficacy time of repaglinide loaded by PLGA-based NPs from 1 day to 1 week with equivalent therapeutic effect<sup>39</sup>. Two types of NPs designated as RPGNP1 and RPGNP2 were prepared in this work, based on PLGA and mPEG-PLGA copolymer respectively. The loading capacity of pure PLGA NPs was  $58.7 \pm 1.3$  and the mPEG-PLGA was less than the former, being  $45.8 \pm 1.2$ . *In vitro* release experiments indicated that both of these two formulations experienced an initial burst release and a following sustained release process. However, different trends were observed in the *in vivo* investigations. The blood glucose level of diabetic rats treated by PLGA NPs decreased within the first 24 h and began to climb afterward, while the hypoglycemic effect of mPEG-PLGA NPs maintained over 7 d (figure 12). A possible reason could be the different affinity to liver macrophages: RPGNP2 were hardly identified by liver macrophages due to their PEG structure and able to retain for a relatively long time, while RPGNP1 could be rapidly identified and neutralized by liver macrophages without the PEG sheltering effect.

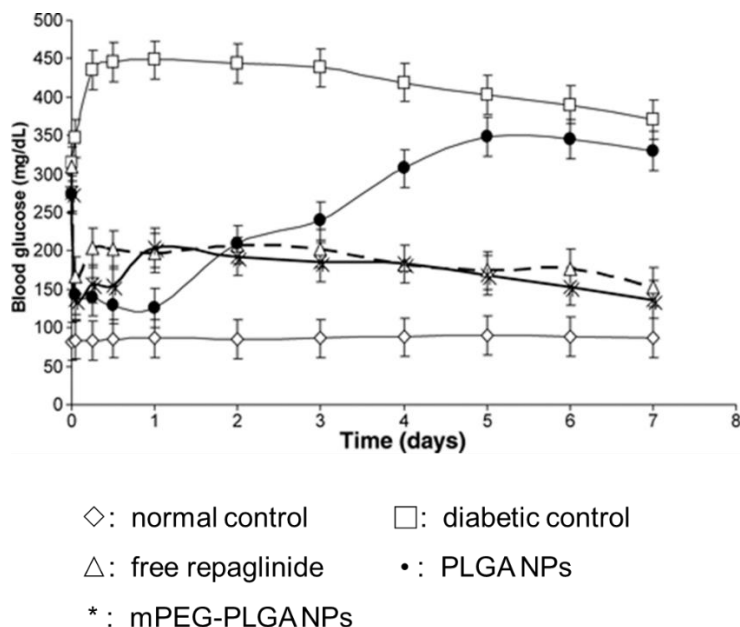


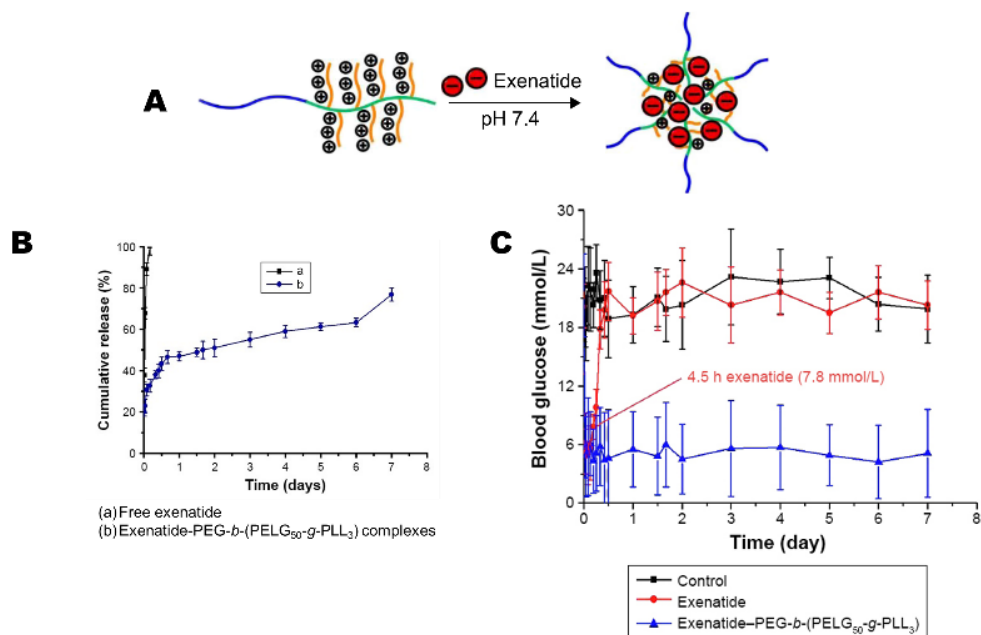
Figure 12. Blood glucose level of different *in vivo* experimental groups in the work of Shelesh JAIN and Swarnlata SARAF.

Yanan Shi *et al.* prepared a different type of PLGA-PEG NPs, with Fc modified for the oral delivery of



404 exenatide<sup>33</sup>. Exenatide is a GLP-1 analogue with 39 amino acids. GLP-1 is a versatile peptide generated by  
405 the proglucagon gene, mainly produced by the intestinal L-cells<sup>97</sup>. GLP-1 have been widely concerned as  
406 incretin hormones to treat diabetes<sup>98,99</sup>. However, a fatal demerit of natural GLP-1 is extremely short  
407 lifespan. Due to the high affinity with plasma dipeptidyl peptidase 4 (DPP-4), GLP-1 can be neutralized  
408 within 2 min<sup>97</sup>. By contrast, the half-life of exenatide is 2.4 h<sup>100</sup>. The Fc-targeted NPs could bind to Fc  
409 receptor which is expressed on the epithelial cells in the small intestine that could help the absorption of  
410 NPs. The *in vitro* and *in vivo* studies verified the better cell uptake and hypoglycemia maintaining  
411 performance. Their group also developed low molecular weight protamine (LMWP)-contained PEG-PLGA  
412 NPs for the oral delivery of Zn-exenatide complexes<sup>38</sup>. In this delivery system, the LMWP could increase  
413 the penetrability of the whole delivery system which was confirmed by the cellular uptake experiment  
414 compared with the pure copolymeric delivery system, the bioavailability also exhibited great improvement.

415 Similar to the strategy of Jun Wang *et al.*, using the electrostatic interaction to connect drugs and  
416 carriers, Fei Tong prepared PEG-*b*-(PELG<sub>50</sub>-*g*-PLL<sub>3</sub>) polymeric NPs to carry exenatide via electrostatic force  
417 between the negative exenatide molecules and positive polymers under pH 7.4 (**figure 13 A**)<sup>37</sup>. As Fei Tong  
418 reported, the loading efficiency of these PEG-*b*-(PELG<sub>50</sub>-*g*-PLL<sub>3</sub>) NPs on exenatide was 12.11%. The  
419 cumulative release profile indicated that the release of exenatide displayed a very stable and sustainable  
420 manor (**figure 13 B**), and the observation of blood glucose level revealed a significantly prolonged  
421 hypoglycemic activity (**figure 13 C**). Besides, alleviated diabetic nephropathy was also observed in this  
422 study.



**Figure 13. A:** formation of exenatide-loaded PEG-b-(PELG<sub>50</sub>-g-PLL<sub>3</sub>) NPs; **B:** cumulative release profile of NPs; **C:** blood glucose concentration of groups treated differently<sup>37</sup>.

In addition to delivery GLP-1 analogues, according to Tianqi Nie *et al.* reported, plasmids DNA encoding GLP-1 were also able to be delivered by NPs<sup>36</sup>. DNA formulations are of various advantages but the rapid degradation of pure DNA and the poor ability to cross the mucus layer in the gastrointestinal tract are still the main challenges in their applications<sup>36,101</sup>. PEI has already been reported widely as an efficient transfection compounds<sup>102-104</sup>. As **figure 14** showed, in the work of Tianqi Nie *et al.*, plasmid DNA encoding GLP-1 was complexed with linear PEI to form NPs. Afterwards, DPPC and 1,2-dimyristoyl-*rac*-glycero-3-mPEG-2000 (DMG-PEG) were adopted to coating NPs, forming linear PEI/plasmid DNA NPs with a hydrophilic and electrostatically neutral shell which could benefit the penetration process of mucus layer since its highly hydrophilic with abundant cationic charges via an exquisite method named as flash nanocomplexation (FNC).

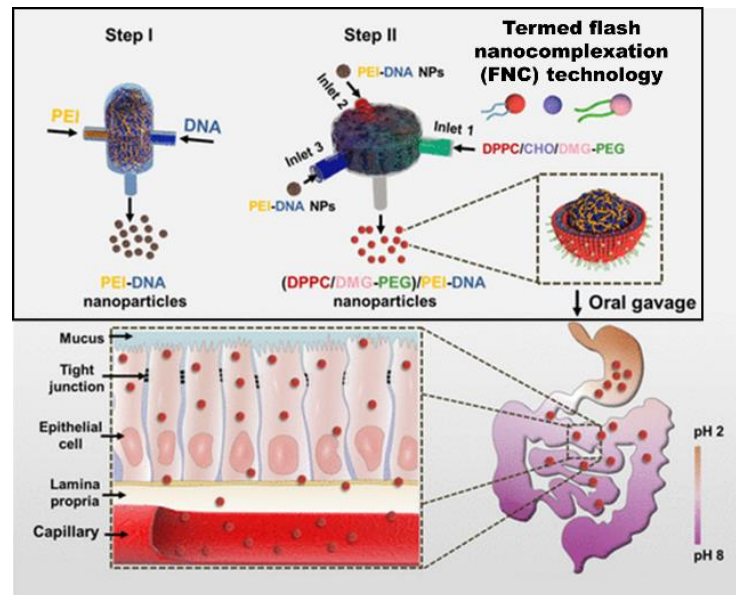


Figure 14. Preparation of linear PEI/plasmid DNA NPs coated by DPPC/DMP-PEG via FNC technology and the expected penetration process through the gastrointestinal tract<sup>36</sup>.

*In vitro* investigations indicated the obviously reduced cytotoxicity induced by the DPPC/DMG-PEG shell and the high transfection efficiency in A549 and HeLa cell lines. *In vivo* experiments pointed out the high efficiency of transfection in the lung and liver, while the good biocompatibility was verified by ameliorated liver damage observed in the toxicity evaluation. Besides, stable expression of GLP-1 resulted in sufficient blood glucose regulation capacity, confirming the potential of the FNC-prepared linear PEI/plasmid DNA NPs coated by DPPC/DMP-PEG.

In order to further improve the physiochemical properties to promote the insulin delivery capacity of PEGylated NPs, pharmaceutical spray drying was adopted in the preparation of NPs (figure 15). Various benefits to drug encapsulation of this formulation technology have been reported<sup>105,106</sup>.

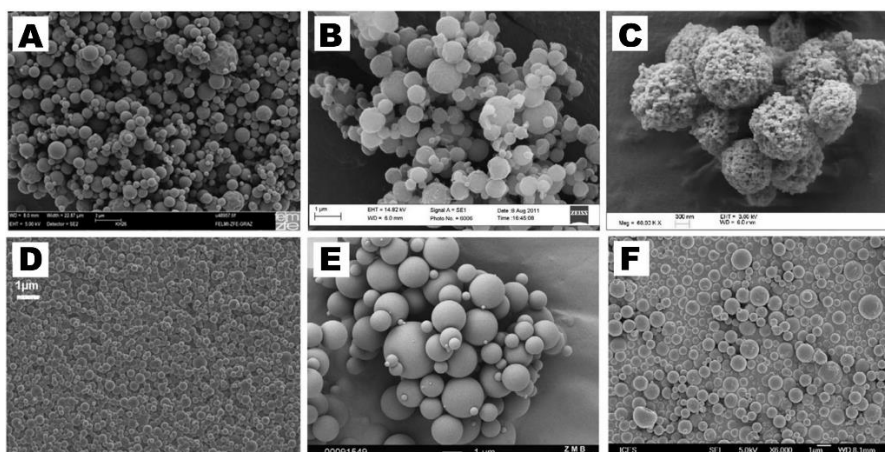


Figure 15. SEM images of several NPs produced by spray drying collected by Cordin Arpagaus *et al.*<sup>106</sup>

452

453 Spray freeze drying (SFD) is deriving from the general spray drying have been applied to prepare NPs  
454 as insulin carriers<sup>107</sup>. In this technology, the notable advantage is it can maximum retain the insulin's  
455 bioactivity due to the drying process under low temperature.

456 Fei Yu *et al.* took the advantage of SFD to form a type of hydroxypropyl methylcellulose  
457 phthalate-coated hard gelatin capsules (HP55) loaded with mono-dispersed microparticles containing  
458 insulin-loaded PLGA-lipid-PEG nanoparticles (designated as micro-particles@INS-PLGA-lipid-PEG NPs) for  
459 oral administration of insulin (**figure 16**), exhibiting excellent entrapment efficiency (92.3%), much more  
460 cellular uptake efficiency than the naked insulin and prolonged decreasing blood glucose level in diabetic  
461 mice with oral administration<sup>35</sup>. This kind of NPs combine the advantages of both polymeric NPs and  
462 liposomes<sup>108</sup>. Three different functional domains constitute the NPs: a hydrophobic PLGA core as the  
463 insulin carrier, an amphiphilic middle layer composed of soybean phosphatidylcholine (SPC) promotes the  
464 delivery efficiency and a PEG shell provides physiological stability. Gradient insulin release and elimination  
465 were observed after oral administration of the prepared LPNs compared with the rapid release rate  
466 induced by subcutaneous injection.

467

468 Sampath Malathi and coworkers prepared a series of D- $\alpha$ -tocopherol PEG 1000 succinate  
469 (TPGS)-emulsified PEG-capped PLGA NPs (ISTPPLG NPs) for insulin delivery via oral administration<sup>32</sup>. The  
470 rat trials suggested that the ISTPPLG NPs could successfully decrease the serum glucose level and last for  
471 24 h. Notably, the ISTPPLG NPs showed a regenerative effect of the liver, kidneys and pancreas on diabetic  
472 rats compared to normal control rats.

473

## 474 **5 PEGylated hydrogels in antidiabetic treatments**

475 Hydrogel is a type of water-swollen networks mainly consisting of polymer<sup>109</sup>, and PEG is a widely used

ingredient for hydrogels which have been extensively studied to achieve controllable and sustainable antidiabetic drug delivery. Plenty of studies report the DDSs based on hydrogels<sup>41-46,110-115</sup>. Basically, a wide range of substrates can be loaded in the PEG-based hydrogel systems due to their unique structural characters: the highly customizable block length of PEG and other components of hydrogel allow the adjustment of equilibrium between hydrophobicity and hydrophilicity, enabling the universality of different molecules<sup>116</sup>. Another advantage of copolymeric hydrogels is the diverse stimuli-responsive capacity. Thermosensitive, pH-responsive hydrogels and many other species have been developed and applied in DDSs<sup>117</sup>. Besides, varying the ratio of different components and combining two or more distinct copolymers also could improve loading capacity. Some examples of diabetes treatment are listed below.

Thermosensitive hydrogels are the most investigated stimuli-sensitive species<sup>118</sup>, already been applied in the delivery of biomacromolecules<sup>119</sup>. Phase diagrams reveal the conversion between liquid and solid (figure 16). The lower critical solution temperature (LCST) is the lowest temperature that the polymer remains soluble in aqueous solvent which is injectable. Once beyond the LCST, polymer will transfer to solid gel state to form local drug storage.

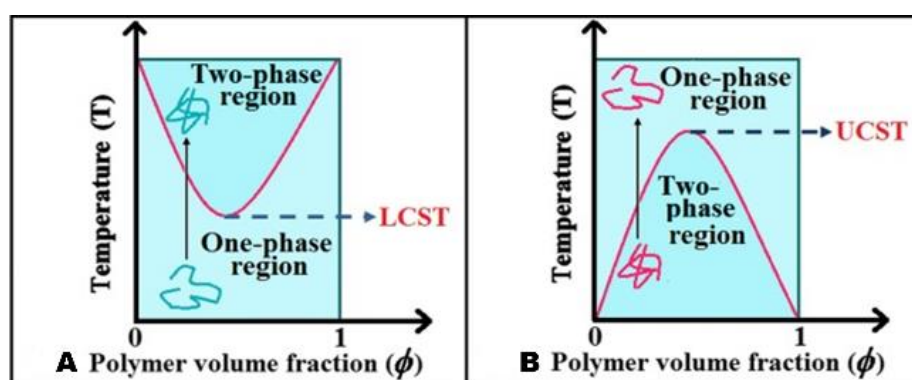
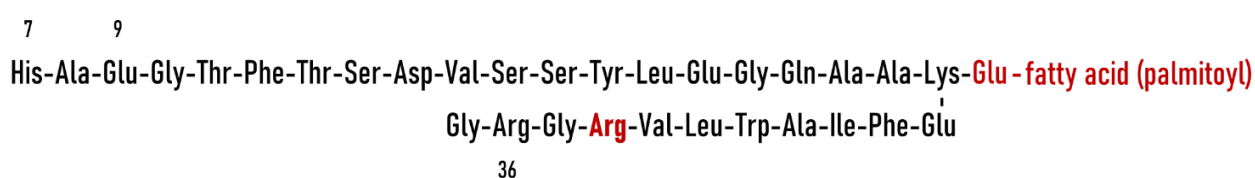


Figure 16. Phase diagrams mark the lower critical solution temperature (LCST) and the upper critical solution temperature (UCST) as the temperature barriers between monophasic and biphasic mixture<sup>117</sup>.

Up to date, thermosensitive hydrogels have already been adopted to deliver antidiabetic drugs. Yipei Chen *et al.* described a series of injectable thermosensitive hydrogels to achieve sustained release of liraglutide (lira)<sup>41</sup>. The hydrogels were composed by poly ( $\epsilon$ -caprolactone-*co*-glycolic acid)-PEG-poly

497 ( $\epsilon$ -caprolactone-*co*-glycolic acid) (PCGA-PEG-PCGA), obtained via typical ROP, being of similar molecular  
498 weights while the ratio of  $\epsilon$ -caprolactone-*co*-glycolide was various. As reported, these biocompatible  
499 copolymers were able to convert to gel from aqueous solution as the ambient temperature increased,  
500 which means that copolymers loaded lira were injectable under room temperature, once these  
501 thermosensitive materials enter the body, they could form stable hydrogels *in situ* to build drug storages  
502 and release pre-encapsulated lira continuously. This copolymer hydrogel formulation fulfills the demand of  
503 both sustain drug delivery and painless administration. Meanwhile, pancreatic function benefits were  
504 observed, indicating the considerable clinical value of these thermosensitive hydrogels.

505 The loaded drug, lira, is a palmitoyl-acylated derivative of GLP-1. As **figure 17** exhibited, the replacement  
506 of lysine with arginine at position 34 and the linkage of a 16-carbon fatty acid at position 26<sup>120</sup> impede the  
507 degradation induced by DPP-4<sup>121</sup>, extending the half-life of lira (11-13 h) compared with the unmodified  
508 GLP-1 (no more than 2 min)<sup>122</sup>. Besides, various pharmacological activities including the normalization of  
509 serum glucose level, the regulation of cardiovascular situation<sup>123</sup> and the promotion of  $\beta$ -cell  
510 proliferation<sup>124</sup> have been reported.



511  
512 **Figure 17.** Structure of liraglutide<sup>125</sup>.  
513

514 A PLGA-PEG-PLGA triblock copolymeric thermosensitive hydrogel was reported<sup>42</sup> as the carrier of  
515 another versatile GLP-1 receptor agonists lixisenatide (lixi)<sup>126-128</sup> since the electronic interaction between  
516 lixi and PLGA-PEG-PLGA benefit the stability of lixi. Except for PLGA-PEG-PLGA, PCGA-PEG-PCGA  
517 copolymers and the mixture of these two copolymeric systems were also investigated. According to the  
518 literature, blending hydrogels are of more stable degradation performance *in vivo* than PLGA-PEG-PLGA or  
519 PCGA-PEG-PCGA hydrogel alone. Most of all, as the pharmacokinetic study of mixture suggesting,

520 remarkable pharmaceutical improvements were observed including prolonged half-life time and enlarged  
521 AUC (table 2).

522 **Table 2.** Pharmacokinetic investigation unfolded the enhancement of hydrogel formulation<sup>42</sup>.

sample	C <sub>max</sub> <sup>a</sup> (ng/mL)	<sup>b</sup> T <sub>max</sub> (h)	<sup>c</sup> t <sub>1/2z</sub> (h)	AUC <sub>(0-last)</sub> (h ng/mL) <sup>d</sup>	MRT <sup>e</sup>
free Lixi	106.4	1.0	2.2	378.9	2.8
Lixi/Gel	24.5	0.5	30.3	2891.6	94.6

523 a: maximum plasma concentration; b: time required to reach the maximum plasma concentration; c: plasma half elimination  
524 time; d: area under the curve; e: mean retention time.

525

526 Divya Sharma and Jagdish Singh prepared a PLGA-PEG-PLGA copolymeric hydrogel system to load  
527 chitosan-zinc-insulin complexes<sup>40</sup>. The complex of insulin and zinc and the application of oleic-modified  
528 chitosan polymer significantly stabilized insulin and its distribution process among this thermosensitive  
529 injectable hydrogel. *In vivo* investigation indicated the more stable blood concentration of insulin and the  
530 prolonged blood glucose regulation time compared with free insulin.

531 Mei Zhang and coworkers<sup>44</sup> reported an oxidation-responsive hydrogel polymerized by  
532 4-arm-PEG20k-SH and H<sub>2</sub>O<sub>2</sub>-breakable diacrylate (figure 18). Fluorescein isothiocyanate (FITC) insulin and  
533 GOx were loaded into the hydrogel network.

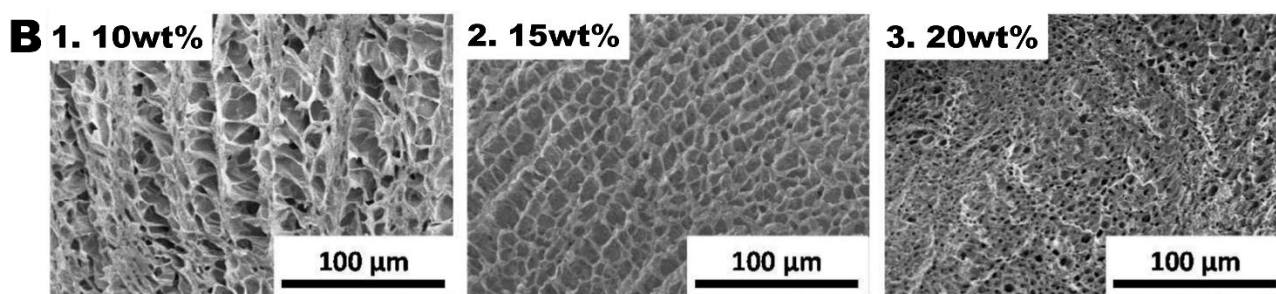
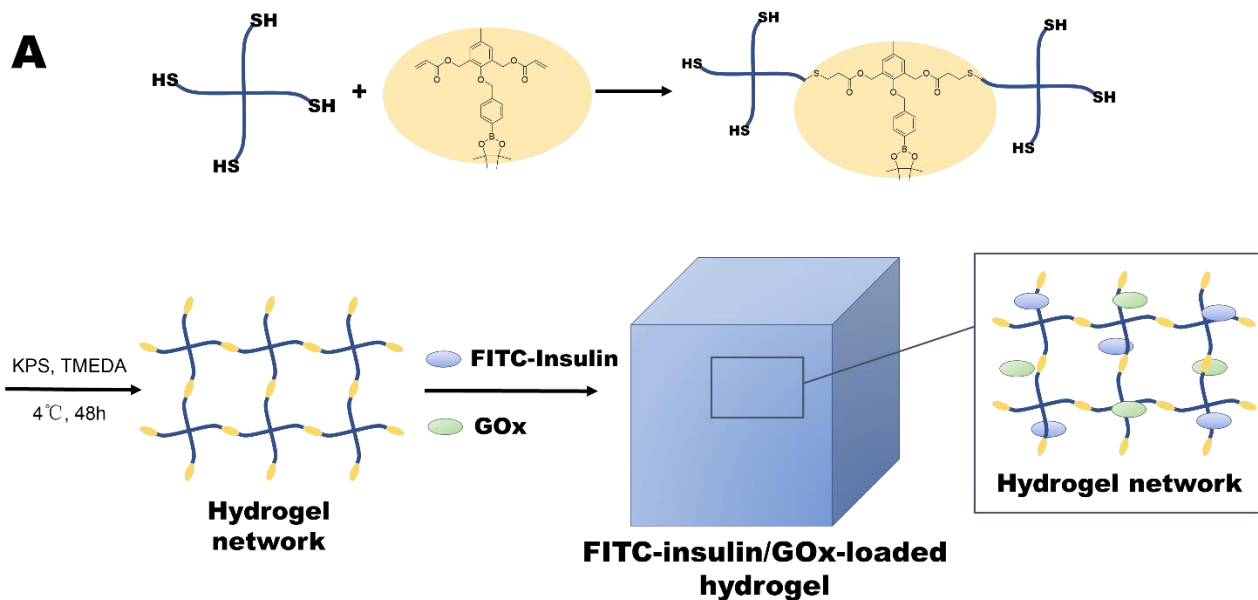


Figure 18. A is the preparation process of FITC-insulin/GOx-loaded oxidation-responsive hydrogel; B are the

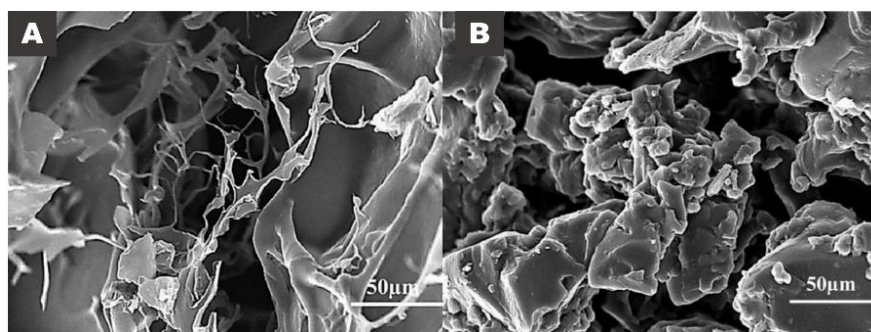
SEM images of the hydrogels prepared by Mei Zhang *et al.*<sup>44</sup>

The acrylic moieties hanging in the end of the main chains provided oxidation-responsive capacity. The degradation process can be classified into two approaches, affording 5 parts including phenylboronic acid, acrylic acid, thioether moiety. Phenylboronic acid was oxidized by  $H_2O_2$ , experiencing the 1,6/1,4-elimination with relatively high speed compared with the rate of thioether oxidation<sup>129</sup>. This discrepancy pointed out that the decomposition of phenylboronic acid played a major role in the degradation of hydrogels.

A number of natural materials are of good biocompatibility with low cost. The integration of synthetic and native substances might be able to overcome the demerits of each other and combine their strengths, affording drug carriers with ideal properties and acceptable prices. One of the typical biodegradable materials being able to combine with synthetic polymers like PEG to form hydrogel is chitosan (CS). CS is a



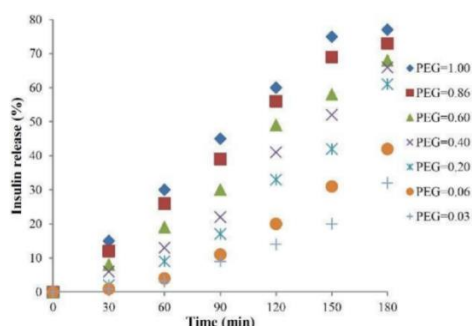
548 natural polysaccharide composed of a series of linear copolymers of D-glucosamine and  
549 N-acetyl-D-glucosamine<sup>130</sup>. Due to properties such as good biocompatibility and encapsulation capacity  
550 of negatively charged proteins and peptides, CS has captured much attention in DDSs<sup>131,132</sup>. However, CS  
551 requires further structural modifications to improve its loading capacity before its application<sup>133-135</sup>.  
552 PEGylation could be considered as an acceptable method to extend blood half-life time<sup>136</sup> and further  
553 enhance the biocompatibility of CS<sup>137</sup>. Bahman Vasheghani Farahani and coworkers fabricated a  
554 glucose-responsive semi IPN hydrogel by free radical cross-linking polymerization of CS, acrylamide (AAM)  
555 and PEG, using hydrogen peroxide as the initiator and N, N'-methylenebisacrylamide (MBA) as the  
556 crosslinker<sup>45</sup>. Figure 19 displays its exact structure.



557  
558 **Figure 19.** SEM image of chitosan semi-IPN hydrogel (A) and insulin loaded chitosan semi-IPN hydrogel (B)<sup>45</sup>.

559

560 The glucose-responsive property was generated mainly by GOx while CAT also contributed. Insulin was  
561 loaded inside the hydrogel by the swelling-diffusion technique. Interestingly, increased PEG ratio led to the  
562 rise of swelling ratio, drug loading capacities and entrapment efficiency. *In vitro* investigation showed that  
563 per 0.1 g of hydrogel released 150 units of insulin under 500 mg/dL of glucose concentration, and the  
564 insulin release rate can be flexibly adjusted by altering the ratio of PEG (figure 20).



565

566

567

**Figure 20.** *In vitro* insulin release profiles of semi IPN hydrogels with different PEG ratio prepared by Bahman Vasheghani Farahani *et al.*<sup>45</sup>

## 569 **6 Conclusion**

570 Diabetes mellitus is a severe chronic metabolic disorder that causes huge economic losses and physical  
571 pain of patients. Numerous DDSs for antidiabetic drugs, including micelles, vesicles, nanoparticles,  
572 microneedles and hydrogels, have been developed to achieve efficient and convenient administration. PEG  
573 is a series of amphiphilic polymers which have been studied elaborately in DDSs. The PEG modified DDSs  
574 allow multiple physiochemical, pharmacokinetic or pharmacodynamic promotions of antidiabetic drugs  
575 such as insulin, GLP-1 analogues and others.

576 The large number of combinations between PEG and other natural or synthetic molecules such as  
577 chitosan and PLGA provide abundant types of carriers with diverse characters to fulfill complex demands  
578 of drugs. Generally, in the micelle systems, PEG functions as the hydrophilic tails to accelerate the  
579 self-assembly of amphiphilic copolymers to afford micelles. Besides, the hydrophilic shell formed by PEG is  
580 able to shelter micelles and loaded molecules from undesirable enzymolysis. In the meantime, the  
581 purposive modified hydrophobic heads of copolymers could endow micelles with stimuli-responsive  
582 capacity, such as the phenylboronic acidified hydrophobic heads could perceive the slight fluctuation of  
583 blood glucose level, to achieve controllable and sustainable drug release. Basically, due to the  
584 sophisticated structures, several ingredients could be loaded inside the PEGylated vesicles simultaneously.  
585 Therefore, vesicles could respond to multiple physiological stimuli to perform a promoted release profile.  
586 The highly improved drug delivery efficiency of PEGylated NPs also has been proven such as extended *in*  
587 *vivo* half-life time. In the designing and preparation of hydrogels, PEG is widely adopted as an ideal  
588 component to build these porous networks. The various combinations between PEG and other natural or  
589 artificial materials provide plenty of strategies to develop DDSs. For instance, the thermosensitive  
590 hydrogels obtained via integrating thermosensitive materials and PEG exhibit significantly extended  
591 internal lifespan.

592 This review highlights the advantages and the versatility of PEG in the designing and preparation of  
593 antidiabetic DDS. Benefits like extended plasma half-life, reduced aggregation, enhanced delivery  
594 efficiency, improved biocompatibility and stimuli-responsive capacity have been summarized. Predictably,  
595 PEG will be continuously studied as an important chemical to develop DDS in the treatment of diabetes  
596 mellitus.

597

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