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Sustained and Controlled Release Injectable Systems: Biodegradable Polymers in Long-Acting Therapeutics

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Abstract
Chronic and complex disease states requiring prolonged pharmacotherapy present significant challenges to patient adherence and therapeutic outcomes when managed with conventional daily or multiple-daily dosing regimens. Long-acting injectable formulations utilizing biodegradable polymers address these limitations by providing sustained drug release over weeks to months following a single administration. This review examines the scientific principles, formulation strategies, and therapeutic applications of biodegradable polymer-based sustained-release injectable systems. Synthetic polymers including poly (lactic-co-glycolic acid) (PLGA), poly(lactic acid) (PLA), poly(caprolactone) (PCL), and polyethylene glycol derivatives are analyzed alongside natural polymers such as chitosan, gelatin, and alginate, with emphasis on degradation kinetics, biocompatibility, and material properties governing release behavior. Formulation approaches encompass preformed microparticles prepared by emulsion-solvent evaporation and spray drying, in situ forming depots utilizing phase inversion or thermosensitive gelation, and implantable solid matrices including biodegradable rods and drug–polymer composites. Release mechanisms integrate diffusion-controlled processes governed by Fickian principles, degradation-controlled release through hydrolytic chain scission exhibiting bulk or surface erosion characteristics, and swelling-mediated transport in hydrogel systems. Therapeutic applications span oncology with leuprolide acetate depots for prostate cancer, infectious diseases including long-acting antiretroviral formulations for HIV prophylaxis and treatment, and chronic disease management exemplified by sustained-release antipsychotics for schizophrenia and diabetes therapies. Comparative evaluation of microspheres, in situ depots, hydrogels, and implants reveals distinct profiles in drug loading capacity, release duration, and clinical translation potential. Challenges including burst release mitigation, manufacturing scale-up, and regulatory pathways are addressed alongside emerging smart biodegradable systems incorporating stimuli-responsive elements for precision-controlled long-acting therapeutics.

Keywords: Sustained Release Injectables, Biodegradable Polymers, Long-Acting Therapeutics, Polymeric Depots, Controlled Drug Release, Parenteral Drug Delivery

1. Introduction

Chronic disease management and long-term pharmacotherapy impose substantial burdens on patients and healthcare systems through requirements for frequent medication administration. Poor adherence to daily dosing regimens remains a primary cause of therapeutic failure, disease progression, and increased healthcare utilization across diverse therapeutic areas including psychiatric disorders, HIV infection, diabetes, and cancer [1]. Each omitted dose reduces drug exposure, potentially compromising efficacy and, in infectious diseases, promoting resistance development.

Long-acting injectable (LAI) formulations address adherence challenges by maintaining therapeutic drug concentrations for extended periods following single administration, reducing dosing frequency from daily to weekly, monthly, or even quarterly intervals [2]. Beyond adherence benefits, these systems provide stable pharmacokinetic profiles that avoid peak-to-trough fluctuations characteristic of daily dosing, potentially reducing concentration-dependent toxicity while maintaining continuous therapeutic effect.

Biodegradable polymers have emerged as the predominant platform for sustained-release injectable systems, offering controlled degradation to biocompatible metabolites that are eliminated from the body without requiring surgical removal [3]. The ability to tailor polymer composition, molecular weight, and architecture enables precise control over release duration from days to months, accommodating diverse therapeutic requirements. Poly(lactic-co-glycolic acid) (PLGA) has achieved regulatory acceptance in numerous commercial products, establishing precedent for polymer-based depot formulations [4].

This review provides a comprehensive examination of biodegradable polymer-based sustained-release injectable systems. The physicochemical properties and degradation behavior of key synthetic and natural polymers are analyzed, followed by evaluation of formulation strategies for generating injectable depots. Mechanisms governing drug release are examined in detail, with emphasis on the interplay between polymer degradation and drug transport. Therapeutic applications across major disease areas are reviewed, and comparative assessment guides rational selection among formulation approaches. Current challenges and emerging technologies are discussed to provide perspective on future directions in long-acting therapeutics.

2. Biodegradable Polymers in Injectable Systems

2.1. Synthetic Polymers

Synthetic biodegradable polymers offer advantages including reproducible manufacturing, tunable degradation kinetics, and established regulatory pathways. Poly(lactic-co-glycolic acid) (PLGA) represents the most extensively investigated and clinically applied biodegradable polymer for long-acting injectables [5]. PLGA is a copolymer of lactic acid and glycolic acid that undergoes hydrolytic degradation to endogenous metabolites, ensuring excellent biocompatibility. Degradation rate and corresponding drug release duration are controlled by the lactide:glycolide ratio, with higher glycolide content accelerating hydrolysis, and polymer molecular weight, with higher molecular weight prolonging degradation. Commercial PLGA-based products including Lupron Depot® (leuprolide acetate) and Risperdal Consta® (risperidone) demonstrate clinical viability across therapeutic areas [6].

Poly(lactic acid) (PLA) exhibits slower degradation than PLGA due to the methyl group on lactic acid creating steric hindrance to hydrolysis. This property makes PLA suitable for ultra-long-acting formulations requiring release over many months [7]. Poly(caprolactone) (PCL) degrades even more slowly through hydrolysis of ester linkages, with complete degradation requiring 2-4 years depending on molecular weight. PCL's rubbery state at body temperature (glass transition temperature -60°C) provides high drug permeability suitable for diffusion-controlled systems [8].

Polyethylene glycol (PEG) is frequently incorporated into block copolymers (PLGA-PEG, PLA-PEG) to modify hydrophilicity, reduce protein adsorption, and enable thermosensitive gelation behavior. PEG-PLGA-PEG triblock copolymers exhibit sol-gel transitions at body temperature, forming in situ depots upon injection [9].

2.2. Natural Polymers

Natural polymers offer inherent biocompatibility and bioactivity but face challenges in batch-to-batch consistency and degradation rate control. Chitosan, derived from chitin deacetylation, is a cationic polysaccharide that undergoes enzymatic degradation by lysozyme [10]. Its mucoadhesive properties and ability to form hydrogels through physical or chemical crosslinking enable applications in injectable depot systems. Degradation rate depends on deacetylation degree and molecular weight, with higher deacetylation slowing degradation.

Gelatin, derived from collagen hydrolysis, retains cell-binding motifs that promote tissue integration while undergoing enzymatic degradation. Chemical crosslinking with glutaraldehyde or genipin extends degradation time from days to weeks, enabling tunable release [11]. Gelatin's sol-gel transition temperature near body temperature allows formulation as injectable systems that gel upon cooling.

Alginate, an anionic polysaccharide from brown algae, forms hydrogels through ionic crosslinking with divalent cations (calcium, barium). The mild gelation conditions enable protein and cell encapsulation, though mechanical weakness and unpredictable degradation limit applications requiring prolonged release [12].

2.3. Polymer Selection Criteria

Polymer selection for long-acting injectables requires systematic evaluation of multiple parameters. Biocompatibility encompasses not only the polymer itself but also degradation products, which must be non-toxic and readily eliminated [13]. PLGA and PLA degradation products (lactic and glycolic acids) enter the tricarboxylic acid cycle, while PCL degradation products are eliminated through renal and biliary routes.

Degradation kinetics must match desired therapeutic duration. For applications requiring 1-3 month release, PLGA with 50:50 to 75:25 lactide:glycolide ratios is appropriate. Six-month to one-year release may require PLA or high-molecular-weight PLGA with high lactide content [14]. Ultra-long-acting systems (≥1 year) may utilize PCL or PLA with high crystallinity.

Mechanical properties influence formulation processing and *in vivo* performance. Polymers with low glass transition temperatures (PCL, low-molecular-weight PLGA) facilitate microparticle preparation but may exhibit deformation during storage. High-molecular-weight polymers enable implant fabrication with sufficient mechanical strength for surgical handling [15].

3. Formulation Strategies for Long-Acting Injectables

3.1. Polymeric Microspheres and Nanospheres

Microsphere-based depots represent the most extensively commercialized platform for long-acting injectables. These systems comprise drug dispersed or dissolved within polymer matrices, typically 10-200 μm in diameter, administered by conventional needle and syringe [16].

Emulsion-solvent evaporation is the predominant manufacturing technique. Oil-in-water (o/w) emulsification involves dissolving polymer and hydrophobic drug in organic solvent (dichloromethane, ethyl acetate), emulsifying in aqueous continuous phase containing stabilizer (polyvinyl alcohol), and evaporating solvent to harden microspheres [17]. Water-in-oil-in-water (w/o/w) double emulsion enables encapsulation of hydrophilic drugs including peptides and proteins. Process parameters including polymer concentration, emulsification speed, and stabilizer type control particle size and drug distribution.

Spray drying offers continuous manufacturing capability and scalability. Polymer and drug dissolved in volatile solvent are atomized into heated chamber, evaporating solvent to form microspheres in single step [18]. This method avoids large aqueous volumes but requires solvent handling and may expose heat-sensitive drugs to elevated temperatures.

Drug encapsulation efficiency depends on drug solubility in continuous phase, polymer–drug interactions, and formulation parameters. Hydrophobic drugs achieve high encapsulation (>80%) in o/w systems, while hydrophilic drugs in w/o/w systems often exhibit lower efficiency (30-60%) due to partitioning into external aqueous phase [19].

3.2. In Situ Forming Depots

In situ forming depots transform from injectable liquid to solid depot upon administration, avoiding complex microsphere manufacturing and enabling administration through smaller-gauge needles [20].

Phase inversion systems utilize water-miscible organic solvents (N-methyl-2-pyrrolidone, dimethyl sulfoxide) containing dissolved polymer and drug. Upon injection into aqueous tissue environment, solvent diffuses out while water penetrates, causing polymer precipitation and drug entrapment within the forming matrix [21]. Eligard® (leuprolide acetate) utilizes this technology with PLGA dissolved in N-methyl-2-pyrrolidone, achieving 1-, 3-, 4-, and 6-month release formulations through polymer selection.

Thermosensitive gels exploit polymers exhibiting lower critical solution temperature (LCST) behavior, remaining liquid below body temperature and gelling upon injection. PLGA-PEG-PLGA triblock copolymers (ReGel®) demonstrate sol-gel transition near 30°C, enabling subcutaneous injection as liquid that forms gel depot at 37°C [9]. Release duration is controlled by polymer concentration, molecular weight, and copolymer composition.

Injectable hydrogels form through physical or chemical crosslinking following injection. Shear-thinning hydrogels, which flow under applied shear and recover structure upon cessation, enable administration through needles followed by rapid re-gelation at the injection site [22].

3.3. Implantable and Injectable Solid Matrices

Solid implants provide precise control over geometry and release characteristics but require larger-bore needles or surgical insertion. Biodegradable rods fabricated by hot-melt extrusion or compression molding offer uniform drug distribution and predictable release [23]. Zoladex® (goserelin acetate) utilizes PLGA-based rods administered subcutaneously via specialized injector, providing monthly or quarterly release for prostate cancer and endometriosis.

Drug–polymer matrices prepared as cylindrical implants or discs can be inserted subcutaneously through trocars. The defined geometry enables mathematical modeling of release kinetics and quality control through dimensional specifications [24].

4. Mechanisms of Sustained and Controlled Drug Release

4.1. Diffusion-Controlled Release

Diffusion-controlled release from polymeric systems follows Fick's laws of diffusion. In matrix systems where drug is uniformly dispersed, release kinetics are described by the Higuchi equation: $Q = \sqrt{[D(2A - Cs)Cst]}$, where Q is cumulative release, D is diffusion coefficient, A is total drug concentration, Cs is drug solubility, and t is time [25]. This square-root-of-time dependence characterizes diffusion from monolithic matrices where drug concentration exceeds solubility.

Reservoir systems with drug core surrounded by rate-controlling polymer membrane achieve zero-order release when drug activity within core remains constant. Membrane thickness and permeability determine release rate, enabling precise temporal control [26].

4.2. Degradation-Controlled Release

Polymer degradation proceeds through hydrolytic cleavage of ester bonds in synthetic polyesters. Bulk erosion, characteristic of PLGA and PLA, involves water penetration throughout the matrix with degradation occurring uniformly. Auto-catalysis by accumulated acidic degradation products accelerates internal degradation, potentially creating hollow structures [27].

Surface erosion occurs when polymer degradation rate exceeds water penetration rate, limiting degradation to device surface. Polyanhydrides and poly(ortho esters) exhibit surface erosion, maintaining constant device dimensions and near-zero-order release as drug is released only from eroding surface [28].

The interplay between diffusion and degradation determines overall release profile. During initial phase, drug near surface diffuses out (burst release). Subsequently, release may be diffusion-controlled until polymer molecular weight decreases sufficiently to enable erosion-controlled release. Mathematical models incorporating time-dependent diffusion coefficients and porosity changes capture this complexity [29].

4.3. Swelling and Osmotic Mechanisms

Hydrophilic polymers including hydrogels swell upon water absorption, increasing mesh size and enabling drug diffusion through expanded network. Swelling kinetics depend on polymer crosslink density and hydrophilicity. For glassy polymers, solvent penetration creates moving boundary between glassy and rubbery regions, with drug release occurring from rubbery region [30].

Osmotic systems utilize osmotically active agents to generate pressure driving drug solution through orifice. While less common in biodegradable systems, osmotic principles can be incorporated into implantable devices requiring precise zero-order release [26].

5. Therapeutic Applications of Long-Acting Injectable Systems

5.1. Oncology

Long-acting injectables have transformed hormone-sensitive cancer therapy. Leuprolide acetate depot formulations utilizing PLGA microspheres (Lupron Depot®) or in situ forming depots (Eligard®) provide sustained suppression of gonadotropin-releasing hormone for prostate cancer treatment [31]. Monthly, three-month, and six-month

formulations enable continuous androgen deprivation without daily injections.
Goserelin implants (Zoladex®) provide monthly or three-month release for prostate and breast cancer. The solid rod design ensures complete bioerosion with predictable release kinetics [23]. Octreotide long-acting release (Sandostatin LAR®) utilizes PLGA microspheres for monthly administration in acromegaly and neuroendocrine tumors.

5.2. Infectious Diseases
Long-acting injectable antiretrovirals represent a paradigm shift in HIV prevention and treatment. Cabotegravir and rilpivirine combination (Cabenuva®) administered monthly or every two months provides complete HIV treatment, eliminating daily oral tablets [32]. The formulations utilize nanoparticle suspensions rather than polymer depots, demonstrating alternative approaches to long-acting delivery. Tuberculosis treatment, requiring six months of daily multidrug therapy, could benefit from long-acting formulations. Rifampicin and isoniazid loaded in PLGA microspheres demonstrate sustained release over weeks in preclinical studies, potentially enabling fully injectable TB regimens [33].

5.3. Chronic Diseases and Psychiatry
Schizophrenia management has been revolutionized by long-acting injectable antipsychotics. Risperidone microspheres (Risperdal Consta®) provide two-week release, while paliperidone palmitate (Invega Sustenna®) utilizing NanoCrystal® technology enables monthly and quarterly dosing [34]. These formulations reduce relapse rates compared to oral antipsychotics by ensuring continuous medication exposure.
Diabetes management increasingly utilizes long-acting injectable peptides. Exenatide extended-release (Bydureon®) employs PLGA microspheres for weekly administration, improving adherence compared to daily glucagon-like peptide-1 agonists [35]. Insulin depot formulations remain under development, with glucose-responsive systems representing the ultimate goal.
Hormonal disorders including endometriosis and precocious puberty utilize leuprolide and histrelin implants providing year-long release, avoiding frequent injections in pediatric populations [36].

6. Comparative Evaluation of Biodegradable Polymer-Based Systems
6.1. Microspheres
Microspheres offer established regulatory pathway, multiple commercial approvals, and flexible dosing through particle suspension in vehicle. Limitations include complex manufacturing requiring aseptic processing, potential for particle aggregation, and reconstitution requirement before administration [37].

6.2. In Situ Depots
In situ forming depots eliminate reconstitution and reduce manufacturing complexity but require organic solvents that may cause injection site reactions. Release kinetics may be less reproducible than preformed microspheres due to dependence on injection site environment [20].

6.3. Hydrogels
Hydrogels provide excellent biocompatibility and protein-friendly aqueous environment but typically achieve shorter release durations (days to weeks) than polyester systems. Mechanical weakness limits applications requiring prolonged structural integrity [22].

6.4. Implants
Implants offer precise release control and zero-order kinetics capability but require larger-bore needles or surgical insertion. Retrievable implants enable treatment interruption if adverse events occur, an advantage over non-retrievable microparticle depots [23].

7. Challenges and Future Perspectives
7.1. Burst Release Control
Initial burst release, where substantial drug fraction releases within first 24 hours, remains a persistent challenge. Strategies to mitigate burst include coating with rate-controlling membranes, optimizing drug distribution within polymer matrix, and post-manufacturing washing to remove surface-associated drug [38].

7.2. Manufacturing Scale-up
Translation from laboratory to commercial scale requires maintaining particle size distribution, drug loading uniformity, and sterility across batch sizes. Continuous manufacturing approaches including microfluidics and advanced spray drying offer potential for improved consistency and reduced cost [39].

7.3. Regulatory Considerations
Regulatory approval requires demonstration of product quality, safety, and efficacy with particular attention to sterility assurance, release profile consistency, and biocompatibility of degradation products. The complex, multi-component nature of depot formulations requires thorough physicochemical characterization and stability studies [2].

7.4. Patient Adherence and Acceptance
While long-acting injectables improve adherence through reduced dosing frequency, some patients exhibit needle phobia or prefer oral administration. Education and shared decision-making are essential for successful implementation. Injection site reactions, though typically mild, may affect acceptance [34].

7.5. Emerging Smart Biodegradable Systems
Next-generation systems incorporate stimuli-responsive elements enabling on-demand release or glucose-responsive insulin delivery. Enzyme-triggered degradation, pH-sensitive polymers, and thermo-responsive hydrogels expand capabilities beyond passive release [40]. Combination products integrating multiple drugs with independent release kinetics enable fixed-dose combinations for complex regimens.

8. Tables

Table 1: Common Biodegradable Polymers Used in Sustained-Release Injectable Systems

Polymer	Chemical Nature	Degradation Mechanism	Typical Release Duration	Approved Applications
PLGA	Copolymer of lactic and glycolic acid	Hydrolytic degradation (bulk erosion)	1 week - 6 months	Lupron Depot®, Risperdal Consta®, Sandostatin LAR®
PLA	Poly(lactic acid)	Hydrolytic degradation (bulk erosion)	3 months - 2 years	Atridox®, various implants
PCL	Poly(caprolactone)	Hydrolytic degradation (slow)	6 months - 3 years	Capronor®, research stage
PLGA-PEG	Block copolymers	Hydrolytic degradation with increased hydrophilicity	Days - weeks	ReGel® research platform
Chitosan	Poly(N-acetylglucosamine)	Enzymatic (lysozyme)	Days - weeks	Wound dressings, research
Gelatin	Denatured collagen	Enzymatic (proteases)	Days - 3 weeks	Research stage
Alginate	Polysaccharide	Ion exchange, dissolution	Days - weeks	Research stage

Table 2: Formulation Techniques for Polymeric Long-Acting Injectable Systems

Technique	Principle	Advantages	Limitations	Suitable Drug Types
Oil-in-water emulsion-solvent evaporation	Polymer dissolved in organic solvent, emulsified in aqueous phase, solvent evaporated	High encapsulation for hydrophobic drugs, established scale-up	Organic solvent use, shear stress	Hydrophobic small molecules
Water-in-oil-in-water double emulsion	Primary aqueous phase with drug emulsified in polymer solution, secondary emulsification	Enables hydrophilic drug encapsulation	Lower encapsulation efficiency, complex process	Peptides, proteins, hydrophilic drugs
Spray drying	Atomization of polymer-drug solution in heated chamber	Single-step, continuous, scalable	Heat exposure, solvent handling	Small molecules, some peptides
Hot-melt extrusion	Polymer-drug melt processed through die	Solvent-free, high drug loading	Thermal degradation risk	Thermally stable drugs
Phase separation (coacervation)	Polymer precipitation induced by non-solvent	Mild conditions, no heat	Complex, residual solvents	Peptides, proteins
In situ forming depots	Polymer dissolved in water-miscible solvent, precipitates upon injection	Simple manufacturing, low viscosity injection	Organic solvent at injection site	Small molecules, peptides

Table 3: Mechanisms of Controlled and Sustained Drug Release in Biodegradable Injectable Systems

Release Mechanism	Governing Kinetics	Polymer Characteristics	Clinical Implications
Fickian diffusion	$Q \propto \sqrt{t}$ (matrix systems); $Q \propto t$ (reservoir)	Rubbery state, $T_g < 37°C$	Predictable release, independent of degradation initially
Degradation-controlled	Dependent on erosion mechanism (bulk vs surface)	Hydrolytically labile bonds, molecular weight	Delayed release onset, auto-catalysis effects
Swelling-controlled	Case II transport, moving boundary	Hydrophilic, crosslinked	Environment-responsive, potential for zero-order
Bulk erosion	Biphasic: diffusion followed by erosion	Amorphous, hydrophilic	Accelerated terminal release
Surface erosion	Near-zero order, constant device dimensions	Hydrophobic, labile bonds only at surface	Constant release rate, predictable device lifetime
Combination (diffusion + degradation)	Time-dependent rate constants	All degradable polymers	Complex but tunable profiles

9. Conclusion

Biodegradable polymer-based sustained-release injectable systems represent a mature yet continuously evolving platform for long-acting therapeutics. PLGA and related polyesters provide tunable degradation kinetics enabling release durations from weeks to months, with established regulatory precedent supporting product development. Microsphere formulations dominate commercial applications, while in situ forming depots and implantable systems offer distinct advantages for specific therapeutic requirements. Release mechanisms integrating diffusion, degradation, and swelling processes provide multiple levers for controlling drug liberation. Therapeutic applications spanning oncology, infectious diseases, and chronic disease management demonstrate broad utility and clinical impact. Continued advances in burst control, manufacturing technology, and smart responsive systems will further expand capabilities, positioning long-acting injectables as essential tools for improving patient adherence and therapeutic outcomes across diverse disease states.

References

1. Kane JM, Kishimoto T, Correll CU. Non-adherence to medication in patients with psychotic disorders: epidemiology, contributing factors and management strategies. World Psychiatry. 2013;12(3):216-226.
2. Park K, Otte A, Sharifi F, Garner J, Skidmore S, Park H. Potential roles of the glass transition temperature of PLGA microparticles in drug release: a review. Int J Pharm. 2021;606:120851.
3. Makadia HK, Siegel SJ. Poly lactic-co-glycolic acid (PLGA) as biodegradable controlled drug delivery carrier. Polymers. 2011;3(3):1377-1397.

4. Wang Y, Qu W, Choi S. FDA's regulatory science program for generic PLA/PLGA-based drug products. Am Pharm Rev. 2016;19(4):1-6.
5. Anderson JM, Shive MS. Biodegradation and biocompatibility of PLA and PLGA microspheres. Adv Drug Deliv Rev. 2012;64 Suppl:72-82.
6. Okada H. One- and three-month release injectable microspheres of the LH-RH superagonist leuprorelin acetate. Adv Drug Deliv Rev. 1997;28(1):43-70.
7. Lasprilla AJR, Martinez GAR, Lunelli BH, Jardini AL, Maciel Filho R. Poly-lactic acid synthesis for application in biomedical devices. Biotechnol Adv. 2012;30(1):321-328.
8. Woodruff MA, Hutmacher DW. The return of a forgotten polymer—Polycaprolactone in the 21st century. Prog Polym Sci. 2010;35(10):1217-1256.
9. Zentner GM, Rathi R, Shih C, *et al*. Biodegradable block copolymers for delivery of proteins and water-insoluble drugs. J Control Release. 2001;72(1-3):203-215.
10. Rinaudo M. Chitin and chitosan: properties and applications. Prog Polym Sci. 2006;31(7):603-632.
11. Young S, Wong M, Tabata Y, Mikos AG. Gelatin as a delivery vehicle for the controlled release of bioactive molecules. J Control Release. 2005;109(1-3):256-274.
12. Lee KY, Mooney DJ. Alginate: properties and biomedical applications. Prog Polym Sci. 2012;37(1):106-126.
13. Kohane DS, Langer R. Biocompatibility and drug delivery systems. Chem Sci. 2010;1(4):441-446.
14. Schliecker G, Schmidt C, Fuchs S, Wombacher R, Kissel T. Hydrolytic degradation of poly(lactide-co-glycolide) films: effect of oligomers on degradation rate and crystallinity. Int J Pharm. 2003;266(1-2):39-49.
15. Middleton JC, Tipton AJ. Synthetic biodegradable polymers as orthopedic devices. Biomaterials. 2000;21(23):2335-2346.
16. Freiberg S, Zhu XX. Polymer microspheres for controlled drug release. Int J Pharm. 2004;282(1-2):1-18.
17. O'Donnell PB, McGinity JW. Preparation of microspheres by the solvent evaporation technique. Adv Drug Deliv Rev. 1997;28(1):25-42.
18. Vehring R. Pharmaceutical particle engineering via spray drying. Pharm Res. 2008;25(5):999-1022.
19. Wischke C, Schwendeman SP. Principles of encapsulating hydrophobic drugs in PLA/PLGA microparticles. Int J Pharm. 2008;364(2):298-327.
20. Kempe S, Mäder K. In situ forming implants: a review. Eur J Pharm Biopharm. 2012;82(1):1-13.
21. Dunn RL, English JP, Cowsar DR, Vanderbilt DP. Biodegradable in-situ forming implants and methods of producing same. United States patent US 5,324,519. 1994.
22. Guvendiren M, Lu HD, Burdick JA. Shear-thinning hydrogels for biomedical applications. Soft Matter. 2012;8(2):260-272.
23. Hutchinson FG, Furr BJA. Biodegradable polymers for the sustained release of peptides. Biochem Soc Trans. 1985;13(3):520-523.
24. Dash AK, Cudworth GC. Therapeutic applications of implantable drug delivery systems. J Pharmacol Toxicol Methods. 1998;40(1):1-12.
25. Higuchi T. Mechanism of sustained-action medication: theoretical analysis of rate of release of solid drugs dispersed in solid matrices. J Pharm Sci. 1963;52(12):1145-1149.
26. Siepmann J, Siepmann F. Mathematical modeling of drug delivery. Int J Pharm. 2008;364(2):328-343.
27. Göpferich A. Mechanisms of polymer degradation and erosion. Biomaterials. 1996;17(2):103-114.
28. Tamada J, Langer R. Erosion kinetics of hydrolytically degradable polymers. Proc Natl Acad Sci U S A. 1993;90(2):552-556.
29. Siepmann J, Göpferich A. Mathematical modeling of bioerodible, polymeric drug delivery systems. Adv Drug Deliv Rev. 2001;48(2-3):229-247.
30. Brazel CS, Peppas NA. Modeling of drug release from swellable polymers. Eur J Pharm Biopharm. 2000;49(1):47-58.
31. Perez-Marrero R, Tyler RC. Leuprolide acetate in the management of prostate cancer. Expert Opin Pharmacother. 2004;5(6):1341-1352.
32. Orkin C, Arasteh K, Górgolas Hernández-Mora M, *et al*. Long-acting cabotegravir and rilpivirine after oral induction for HIV-1 infection. N Engl J Med. 2020;382(12):1124-1135.
33. Dutt M, Khuller GK. Sustained release of isoniazid from a single injectable dose of poly (DL-lactide-co-glycolide) microparticles. J Antimicrob Chemother. 2001;47(6):829-835.
34. Citrome L. Paliperidone palmitate: a once-monthly injectable antipsychotic for the treatment of schizophrenia. Drugs Today (Barc). 2010;46(5):315-332.
35. DeYoung MB, MacConell L, Sarin V, Trautmann M, Herbert P. Encapsulation of exenatide in poly-(D,L-lactide-co-glycolide) microspheres produced an investigational long-acting once-weekly formulation for type 2 diabetes. Diabetes Technol Ther. 2011;13(11):1145-1154.
36. Eugster EA, Clarke W, Kletter GB, *et al*. Efficacy and safety of histrelin subdermal implant in children with central precocious puberty. J Clin Endocrinol Metab. 2007;92(4):1197-1201.
37. Wischke C, Schwendeman SP. Principles of encapsulating hydrophobic drugs in PLA/PLGA microparticles. Int J Pharm. 2008;364(2):298-327.
38. Huang X, Brazel CS. On the importance and mechanisms of burst release in matrix-controlled drug delivery systems. J Control Release. 2001;73(2-3):121-136.
39. Xu Q, Hashimoto M, Dang TT, *et al*. Preparation of monodisperse biodegradable polymer microparticles using a microfluidic flow-focusing device for controlled drug delivery. Small. 2009;5(13):1575-1581.
40. Al-Tahami K, Singh J. Smart polymer based delivery systems for peptides and proteins. Recent Pat Drug Deliv Formul. 2007;1(1):65-71.

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