

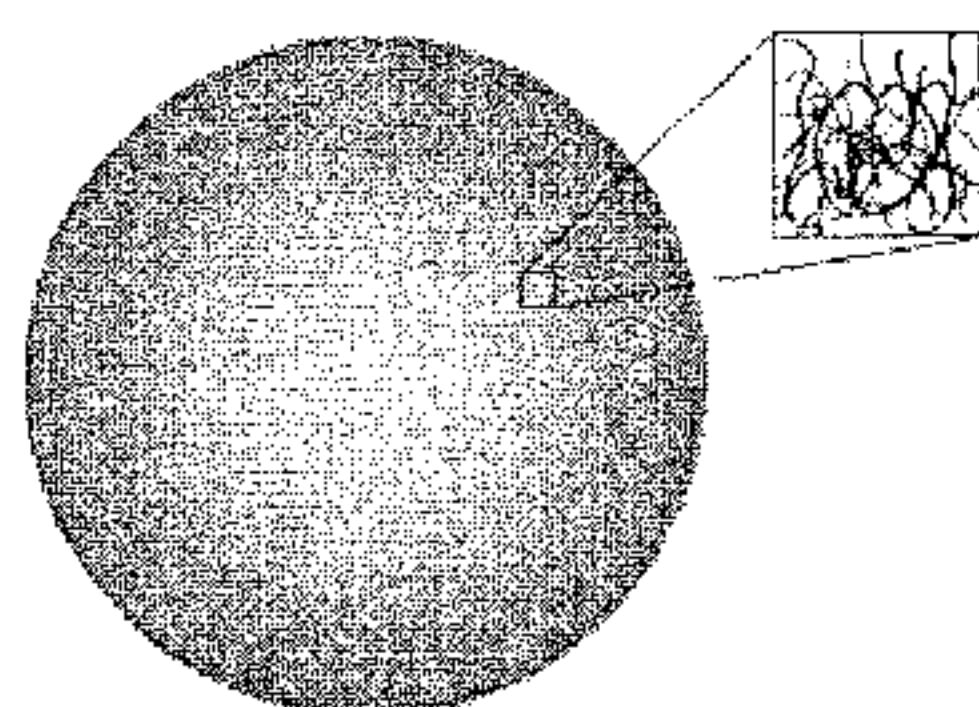


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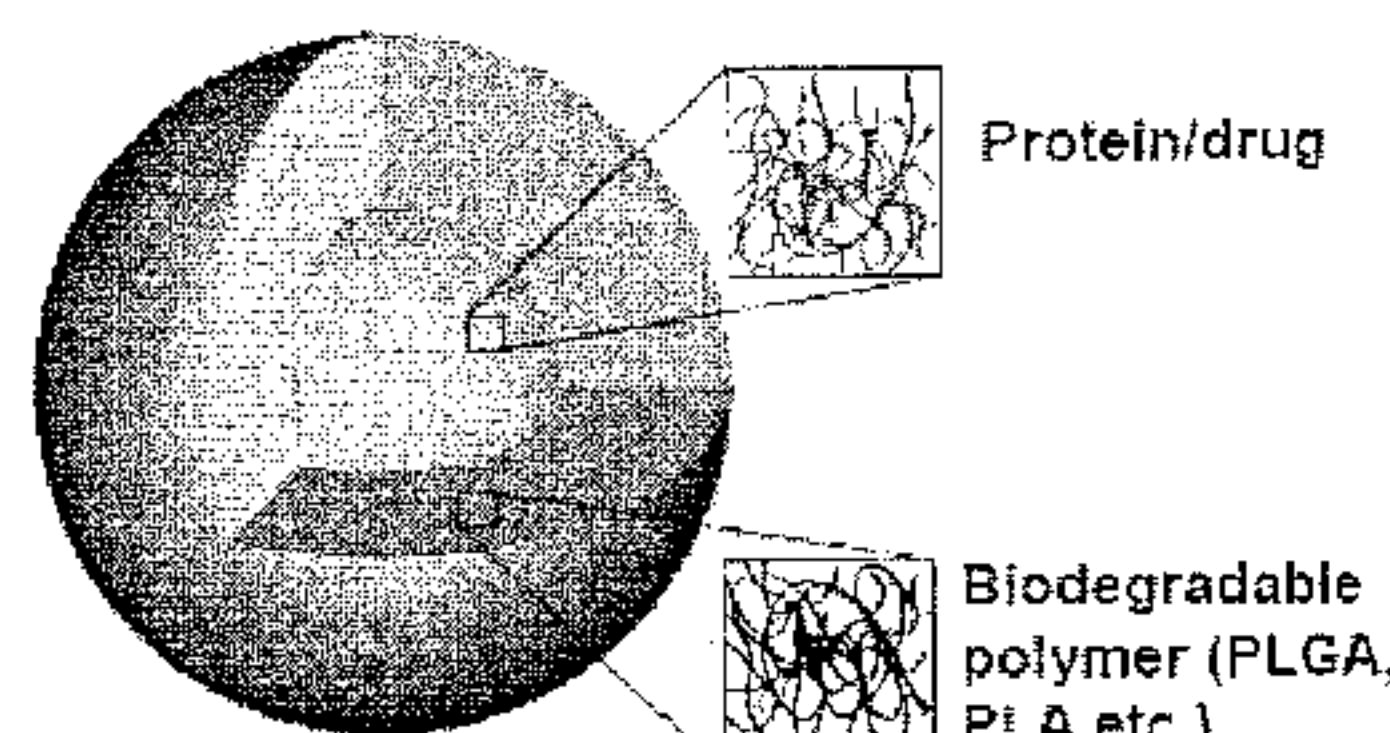
(54) **Titre : FORMULATIONS DE VACCIN MICRO-MOULEES OU IMPRIMEES EN TROIS DIMENSIONS (3D) A LIBERATION PULSATILE**

(54) **Title: MICROMOLDED OR 3-D PRINTED PULSATILE RELEASE VACCINE FORMULATIONS**



Emulsion method

FIG. 11A



3D printing

FIG. 11B

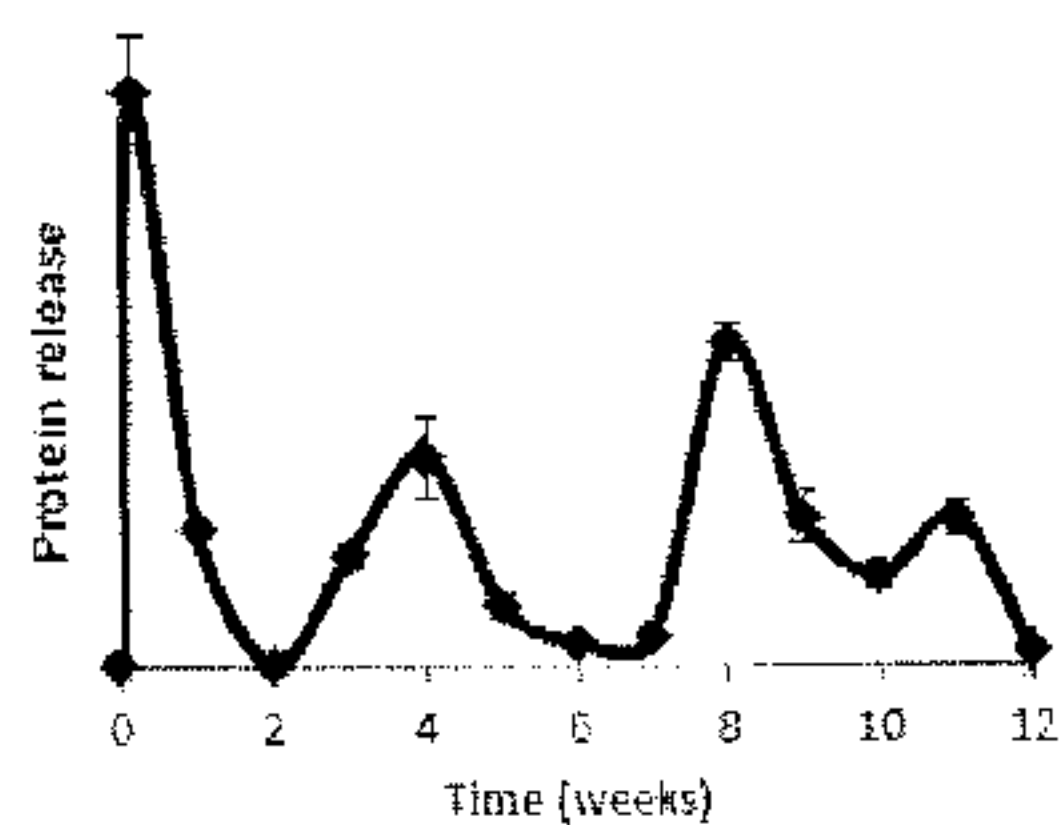


FIG. 11C

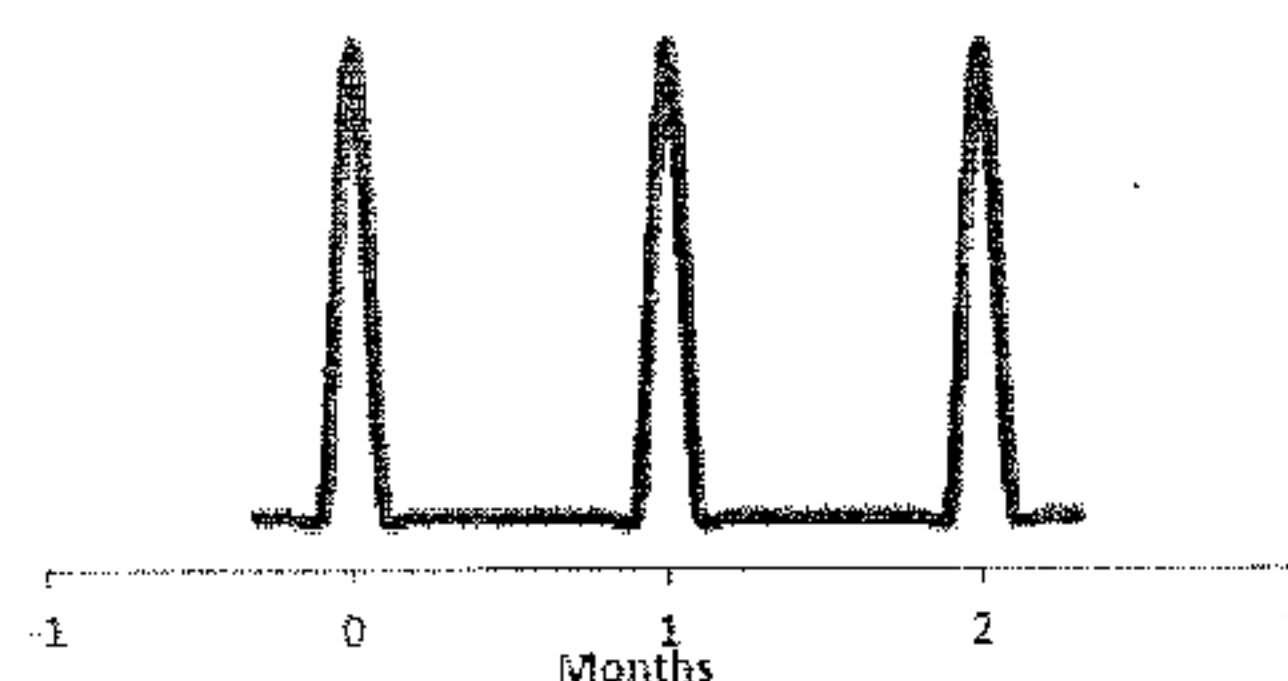


FIG. 11D

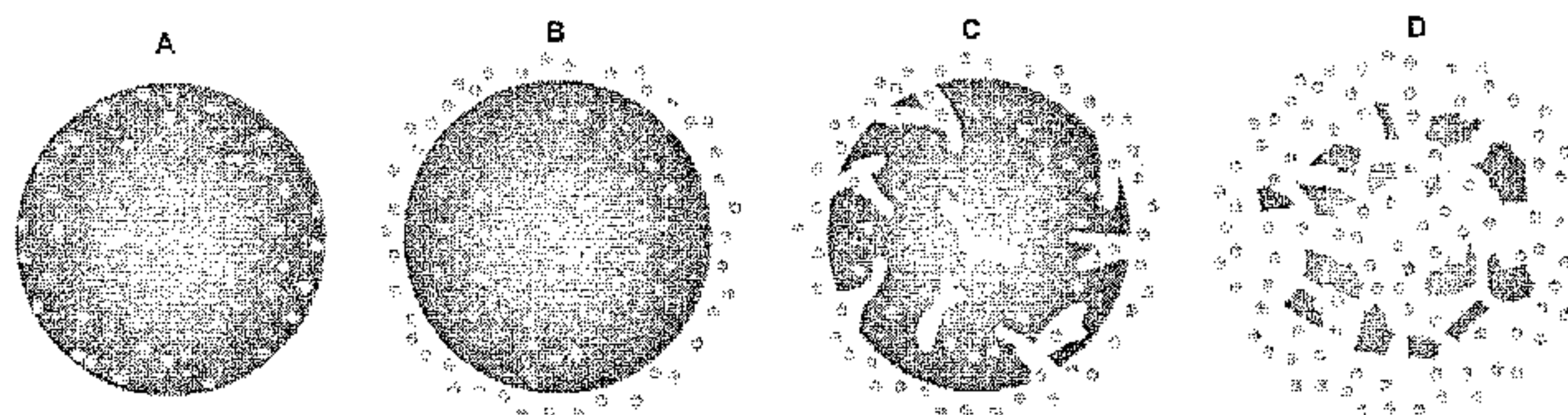


FIG. 11E

(57) **Abrégé/Abstract:**

Emulsion-based and micromolded ("MM") or three dimensional printed ("3DP") polymeric formulations for single injection of antigen, preferably releasing at two or more time periods, have been developed. Formulations are preferably formed of

**(57) Abrégé(suite)/Abstract(continued):**

biocompatible, biodegradable polymers. Discrete regions encapsulating antigen, alone or in combination with other antigens, adjuvants, stabilizers, and release modifiers, are present in the formulations. Antigen is preferably present in excipient at the time of administration, or on the surface of the formulation, for immediate release, and incorporated within the formulation for release at ten to 45 days after initial release of antigen, optionally at ten to 90 day intervals for release of antigen in one or more additional time periods. Antigen may be stabilized through the use of stabilizing agents such as trehalose glass. In a preferred embodiment for immunization against polio, antigen is released at the time of administration, and two, four and six months thereafter.

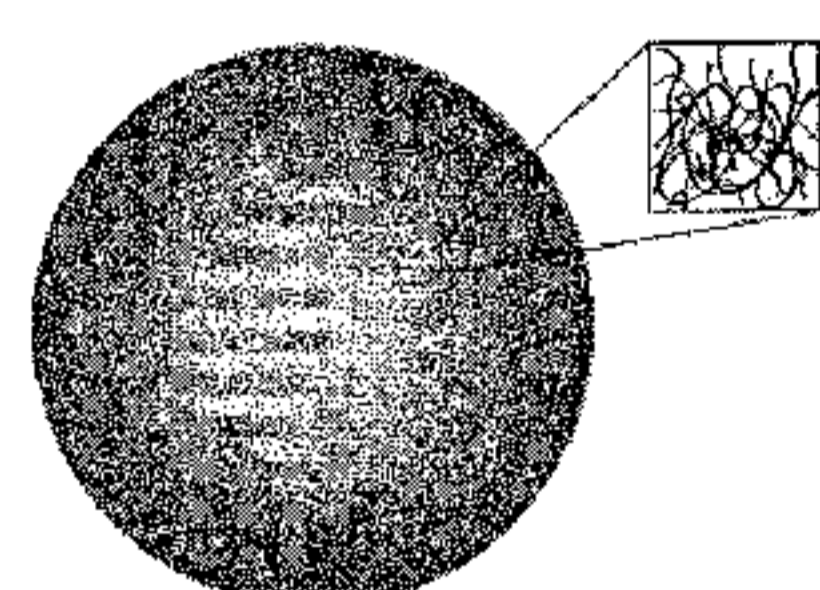
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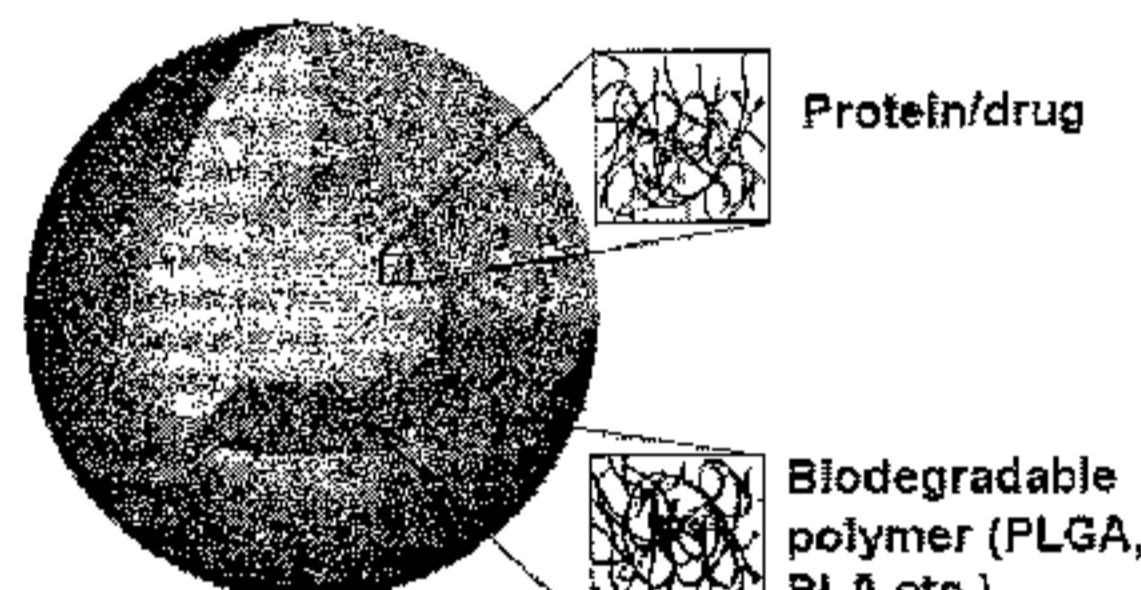
[Continued on next page]

(54) Title: MICROMOLDED OR 3-D PRINTED PULSATILE RELEASE VACCINE FORMULATIONS



Emulsion method

FIG. 11A



3D printing

FIG. 11B

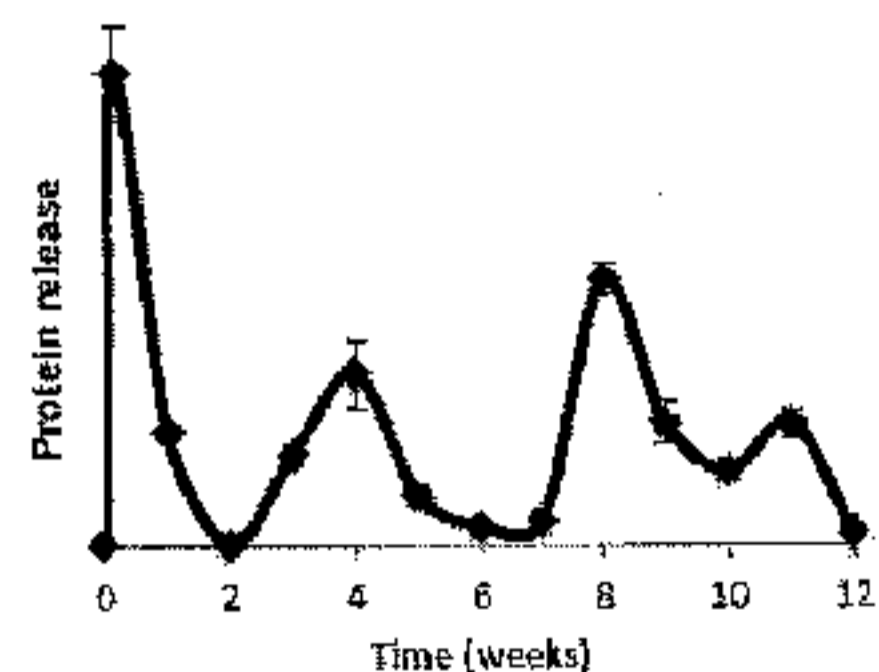


FIG. 11C

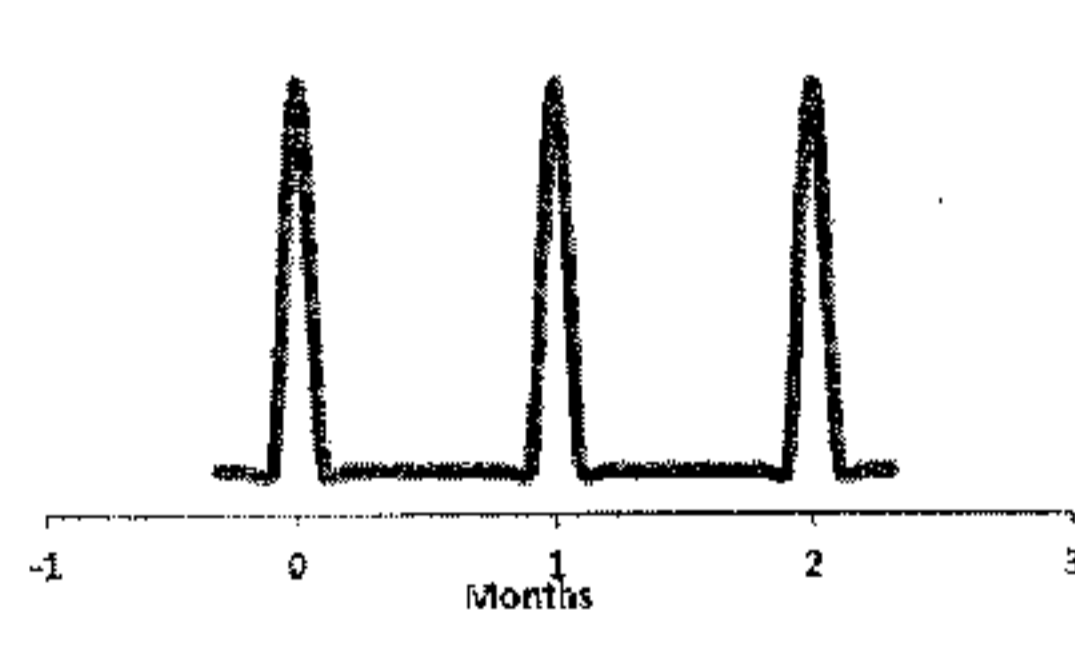


FIG. 11D

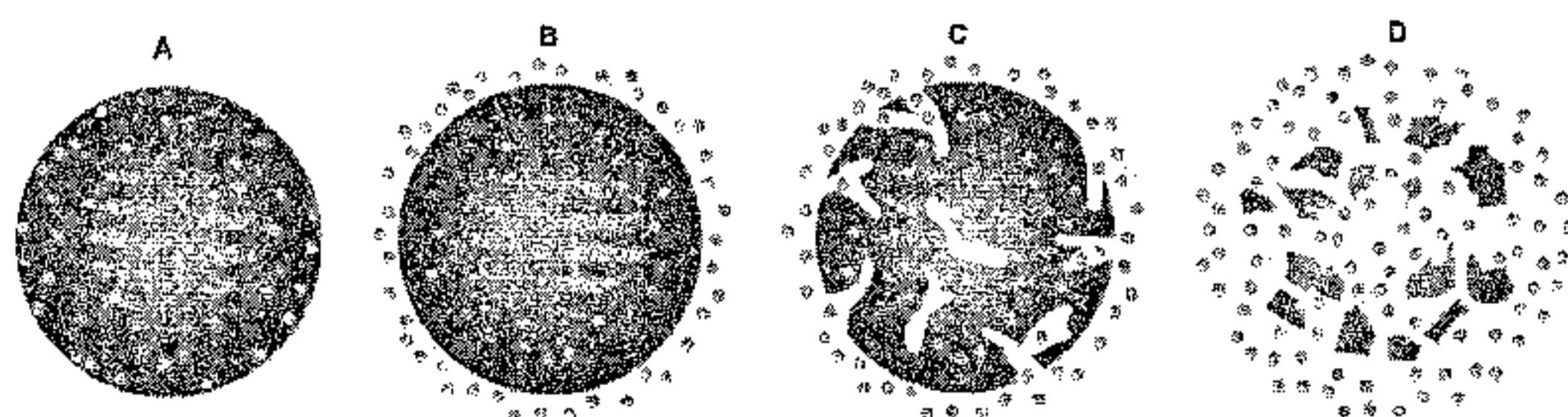


FIG. 11E

(57) Abstract: Emulsion-based and micromolded ("MM") or three dimensional printed ("3DP") polymeric formulations for single injection of antigen, preferably releasing at two or more time periods, have been developed. Formulations are preferably formed of biocompatible, biodegradable polymers. Discrete regions encapsulating antigen, alone or in combination with other antigens, adjuvants, stabilizers, and release modifiers, are present in the formulations. Antigen is preferably present in excipient at the time of administration, or on the surface of the formulation, for immediate release, and incorporated within the formulation for release at ten to 45 days after initial release of antigen, optionally at ten to 90 day intervals for release of antigen in one or more additional time periods. Antigen may be stabilized through the use of stabilizing agents such as trehalose glass. In a preferred embodiment for immunization against polio, antigen is released at the time of administration, and two, four and six months thereafter.

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**Declarations under Rule 4.17:**

- *as to applicant's entitlement to apply for and be granted a patent (Rule 4.17(ii))*
- *as to the applicant's entitlement to claim the priority of the earlier application (Rule 4.17(iii))*

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## MICROMOLDED OR 3-D PRINTED PULSATILE RELEASE VACCINE FORMULATIONS

### FIELD OF THE INVENTION

This invention is generally in the field of injectable vaccine  
5 formulations providing multiple releases of vaccine.

### BACKGROUND OF THE INVENTION

Vaccines typically involve an initial dose of antigen, followed by one  
or more booster doses at defined times after the initial administration,  
typically ten to 60 days later. The need for administration of a booster dose  
10 clearly limits the practicality of vaccines in much of the world, as well as  
increases costs and difficulties in agricultural applications.

Polymeric microspheres have the potential to be effective vaccine  
delivery vehicles. They have the ability to enhance targeting of antigen  
presenting cells (APCs) and have the potential for controlled, sustained  
15 release of antigen-thereby potentially eliminating the need for multiple  
vaccination doses. Further, the polymer matrix can act as a shield from a  
hostile external environment and has the potential to reduce adverse  
reactions and abrogate problems caused by the vaccine strain in  
immunocompromised individuals. PLGA microspheres have been developed  
20 for single immunization, with and without burst release. Given the  
biodegradable nature and sustained release properties that PLGA offers,  
microspheres formulated from PLGA could be useful for the delivery of  
vaccines. Summarized in Kirby *et al.*, Chapter 13: Formation and  
Characterisation of polylactide-co-galactide PLGA microspheres (2013).  
25 PLGA based microparticles are traditionally produced by double emulsion-  
solvent evaporation, nano- precipitation, cross-flow filtration, salting-out  
techniques, emulsion-diffusion methods, jet milling, and spray drying.  
Summarized in Kirby *et al.*, Chapter 13: Formation and Characterisation of  
polylactide-co-galactide PLGA microspheres (2013). PLGA microspheres  
30 can also be formulated to incorporate a range of moieties, including drugs  
and proteins, that can act as adjuvants. It has been contemplated that PGLA

particles produced by these methods can be lyophilized and stored for later use and delivery.

Hanes *et al.*, *Adv. Drug. Del. Rev.*, 28: 97-119 (1997), report on attempts to make polymeric microspheres to deliver subunit protein and peptide antigens in their native form in a continuous or pulsatile fashion for periods of weeks to months with reliable and reproducible kinetics, to obviate the need for booster immunizations. Microspheres have potential as carriers for oral vaccine delivery due to their protective effects on encapsulated antigens and their ability to be taken up by the Peyer's patches in the intestine. The potency of these optimal depot formulations for antigen may be enhanced by the co-delivery of vaccine adjuvants, including cytokines, that are either entrapped in the polymer matrix or, alternatively, incorporated into the backbone of the polymer itself and released concomitantly with antigen as the polymer degrades.

As reported by Cleland *et al.*, *J. Controlled Rel.* 47(2):135-150 (1997), the administration of a subunit vaccine (*e.g.*, gp120) for acquired immunodeficiency syndrome (AIDS) can be facilitated by a single shot vaccine that mimics repeated immunizations. Poly(lactic-co-glycolic acid) (PLGA) microspheres were made that provide a pulsatile release of gp120. Microspheres were made using a water-in-oil-in-water microencapsulation process with either methylene chloride or ethyl acetate as the polymer solvent. The protein was released under physiological conditions in two discrete phases: an initial burst released over the first day and after several weeks or months, a second burst of protein was released. The second burst of protein was dependent upon the PLGA inherent viscosity and lactide/glycolide ratio (bulk erosion).

These studies demonstrate that it is possible to achieve a vaccine response using injectable microparticles. However, no such product has ever been approved for human or animal use. It is difficult to achieve effective loading of antigen, uniformity of encapsulation and release, and extremely low levels of solvent not affecting antigenicity.

It is estimated that precluding the need for a "cold chain" for vaccine distribution through the development of thermo-stable formulations could

save about \$200 million annually. Trouble with implementing these strategies rests on the lack of appropriate cryoprotectant methods. A Summarized in Kirby *et al.*, Chapter 13: Formation and Characterisation of polylactide-co-galactide PLGA microspheres (2013). Similar information and disclosure on nanovaccines can be found in Gregory *et al.*, *Frontiers in Cell and Infect. Microbio.* 3:Article 13 (2013). Nandedkar, *J. Biosci.* 34:995-1003 (2009). Stabilization of proteins included in microspheres is problematic. A number of types of stabilizing excipients have been studied. Summarized in Kim and Pack, *BioMEMS and Biomedical Nanotechnology.* 1:19-50 (2006). Additionally, the type of polymer used for microsphere fabrication, its degradation rate, acidity of the degradation products, hydrophobicity, etc., can also impact the stability of incorporated proteins.

It is therefore an object of the present invention to provide injectable polymeric formulations providing release of encapsulated antigen at two or more times.

It is a further object of the present invention to provide injectable polymeric formulations which do not damage and which can stabilize encapsulated antigen.

It is a still further object of the present invention to provide methods and materials for micromolding and three-dimensional printing of injectable polymeric formulations providing release of encapsulated antigen at two or more times, and the resulting formulations.

### SUMMARY OF THE INVENTION

Emulsion-based and Micromolded (“MM”) or three dimensional printed (“3DP”) polymeric formulations for single injection of antigen, preferably releasing at two or more time periods, have been developed. Formulations are preferably formed of biocompatible, biodegradable polymers. Discrete regions encapsulating antigen, alone or in combination with other antigens, adjuvants, stabilizers, and release modifiers, are present in the formulations. Antigen is preferably present in excipient at the time of administration, or on the surface of the formulation, for immediate release, and incorporated within the formulation for release at ten to 45 days after initial release of antigen, optionally at ten to 90 day intervals for release of

antigen in one or more additional time periods. Antigen may be stabilized through the use of stabilizing agents such as trehalose glass. In a preferred embodiment for immunization against polio, antigen is released at the time of administration, and two, four and six months thereafter. In a preferred  
5 embodiment, leakage between bursts of release is minimal and release occurs over a narrow time frame.

Studies demonstrate the selection of polymer and solvent systems that provides for discrete release of antigen, without overlap, with minimal degradation or damage to the encapsulated antigen. Preferred solvents  
10 include methylene chloride and chloroform, and preferred polymers are polylactic acid ("PLA"), polyglycolic acid ("PGA"), and copolymers thereof ("PLGA").

Formulations are designed for subcutaneous or intramuscular injection via needle or cannula, for topical injection to a mucosal region such as intranasal, or by scarification to the epidermis. Preferred applications are  
15 for administration of antigen eliciting an effective immune response to infectious agents such as bacteria, virus, protozoan and parasitic organisms. However, formulations may also be used for administration of other therapeutic, prophylactic or diagnostic agents, alone or in combination with  
20 antigen.

### **BRIEF DESCRIPTION OF THE DRAWINGS**

Figure 1 is a bar graph showing the effect of lowering excipient:vaccine ratio on percentage of D-antigen retained after drying 26X Trivalent IPV on PLA for 16 hours at room temperature and humidity.

25 Excipients: 10% sorbitol, 8.5% MSG, 8.5% MgCl<sub>2</sub>.

Figure 2 is a line graph showing long-term stability of lyophilized IPV (type I, type II, and type III) with excipients at 37°C in humidified atmosphere. The stability is expressed as percent recover (% Recovery) over time (days). On the line graph, the line for type I IPV is designated (1), the  
30 line for type II IPV is designated (2), and the line for type III IPV is designated (3).

Figures 3A and 3B are graphs of the release of a model protein, bovine serum albumin ("BSA") (Figure 3A, micrograms/10 mg of



microspheres; Figure 3B, % BSA) over time (weeks) for 5% BSA and 0.5% BSA from PLLA, 50 kD.

Figures 4A and 4B are graphs of the release of a model protein, bovine serum albumin ("BSA") (Figure 4A, micrograms/10 mg of  
5 microspheres; Figure 4B, % BSA) over time (weeks) for 5% BSA and 0.5% BSA from PLLA, 100 kD.

Figures 5A and 5B are graphs of the release of a model protein, bovine serum albumin ("BSA") (Figure 5A, micrograms/10 mg of  
10 microspheres; Figure 5B, % BSA) over time (weeks) for 5% BSA and 0.5% BSA from PLLA, 300 kD.

Figures 6A and 6B are graphs of the release of a model protein, bovine serum albumin ("BSA") (Figure 6A, micrograms/10 mg of  
microspheres; Figure 6B, % BSA) over time (weeks) for 5% BSA and 0.5% BSA from P(d,l)LA, 20 kD.

15 Figures 7A and 7B are graphs of protein release over time (weeks) for PLGA formulations (Figure 7A) and bolus injections (Figure 7B).

Figures 8A and 8B are graphs of the release of 0.5% of a model protein, bovine serum albumin ("BSA") compared to release of 0.5% of another model protein ovalbumin ("OVA") (Figure 8A, micrograms/10 mg  
20 of microspheres; Figure 8B, % BSA) over time (weeks) from PLGA (50:50), 20 kD, derivatized with a carboxylic group.

Figures 9A and 9B are graphs of the release of 5% of a model protein, bovine serum albumin ("BSA") compared to release of 5% of another model protein ovalbumin ("OVA") (Figure 9A, micrograms/10 mg  
25 of microspheres; Figure 9B, % BSA) over time (weeks) from PLGA (50:50), 31 kD, derivatized with a carboxylic group.

Figures 10A and 10B are graphs of the release of 0.5% of a model protein, bovine serum albumin ("BSA") compared to release of 0.5% of another model protein ovalbumin ("OVA") (Figure 10A, micrograms/10 mg  
30 of microspheres; Figure 10B, % BSA) over time (weeks) from PLGA (50:50), 31 kD, derivatized with a carboxylic group.

Figures 11A and 11B are schematics of particles made by emulsion (Figure 11A) and by 3D printing or micromolding (Figure 11B), and the

release of protein from the emulsion particle (Figure 11C) and the 3DP particle (Figure 11D), which models the release of protein from bolus injections. Figure 11E is a schematic of polymer degradation-based bursts with one month lag to simulate bolus injections.

5           Figures 12A-12D are schematics of the 3D printing process: Creating the structure of PLGA particles (Figure 12A), filling drugs or proteins into the particles (Figure 12B), drying the drugs or proteins (Figure 12C), and encapsulation of the particles (Figure 12D).

10           Figure 13 is a schematic of the waveform parameters to be optimized to produce uniform single droplets of “ink” (polymer solution) during printing from a piezoelectric nozzle jet.

            Figures 14A-14C are graphs of the percent D-antigen (IPV type I, II, and III) retained after lyophilizing with sugar excipients 1 M trehalose, 1 M sucrose, and 3 M sucrose, then incubating at 4°C, 25°C or 37°C.

15           Figures 15A and 15B are graphs of the percent D-antigen (IPV type I, II, and III) retained after lyophilizing with 0, 0.5, 0.75, 1 or 1.25 M trehalose, then incubating at 4°C (Figure 15A) or 25°C (Figure 15B).

20           Figures 16A and 16B are graphs of the percent D-antigen (IPV type I, II, and III) retained after mixing with solvents until solvent evaporation, with solvents tetrafluoroethylene (“TFE”), dichloromethane (“DCM”), or TFE and DCM, without sugar excipient (Figure 16A) or with 0.75 M trehalose as excipient (Figure 16B).

25           Figure 17A is a graph of the percent D-antigen (IPV type I, II, and III) retained after IVP with 0 M sugar, 0.5 M trehalose, or 0.5 M trehalose-sucrose is printed onto 3D-printed PLA substrate and dried at 25°C, 22% RH. Figure 17B is a graph of the percent D-antigen (IPV type I, II, and III) retained after IVP with 0.25 M trehalose or 0.5 M trehalose-sucrose is printed onto 3D-printed PLA substrate and dried at 25°C, 10.7% RH.

30           Figures 18A and 18B are graphs of the percent D-antigen retained after lyophilization with 0.5 or 0.75 M trehalose then incubating with 25°C (Figure 18A) or after lyophilization with 0.5 M trehalose-sucrose, pipetted, or 0.5 M trehalose-sucrose, jetted, then drying at 25°C, 10.7% RH (Figure 18B).

Figures 19A-19C are graphs showing anti-BSA IgG (antibody) titers (in log<sub>2</sub> values) plotted for the groups presented in Table 3 at 1 week (Figure 19A), 2 weeks (Figure 19B), and 4 weeks (Figure 19C) post-injection. The negative controls are all zero and the microsphere formulations are  
5 generating an equivalent or stronger response compared to the bolus control. (F4 and F8 are blank microspheres, no drug; see Table 4).

Figures 20A-20C are histograms showing size distribution of microspheres prepared with formulation C (Figure 20A), formulation G (Figure 20B), and formulation E (Figure 20C).

10 Figures 20D-20F are histograms showing volume distribution of formulation C (Figure 20D), formulation G (Figure 20E), and formulation E (Figure 20F).

Figures 21A-21D are line graphs showing weekly BSA release (in percentages (%)) from microspheres formulated with 7-17 kDa, 50:50 PLGA (Figure 21A), 24-38 kDa, 50:50 PLGA (Figure 21B), 38-54 kDa, 50:50 PLGA (Figure 21C), and 4-15 kDa, 75:25 PLGA (Figure 21D) containing  
15 0.5, 3, or 5% BSA by weight. Error bars represent standard deviation.

Figures 22A and 22B are line graphs showing IgG antibody titers in mouse sera against BSA released from microparticles and plotted as  
20 geometric mean over time on a log<sub>2</sub> scale. Figure 22A shows low-dose formulations C & G compared to a series of three dose-matched bolus BSA injections. Figure 22B shows high-dose formulation E compared to its dose-matched bolus control. Bolus BSA was injected at 0, 4, and 8 weeks in the control groups. Filled symbols indicate significant differences between the  
25 group with unfilled symbols of the same shape and dose-matched control at that time point determined using ANOVA analysis for A (3 groups) and Student's t-test for B (2 groups). One, two, and three filled symbols indicate p<0.05, p<0.01, and p<0.001 respectively and error bars represent standard deviation.

30 Figures 23A and 23B are dot plots showing peak titers induced by immunization at low (Figure 23A), and high (Figure 23B) BSA dosing. Peak titers for all microparticle formulations occurred at week 4 while both bolus injection groups peaked at week 10. NS represents not significant when

using a Student's t-test with Holm-Bonferroni correction for multiple comparisons at a significance level of 0.05. Adjusted p values for formulations C, G, and E are 0.0645, 0.0543, and 0.0784 respectively.

Figures 24A-24C are schematics of the micromolding process showing the steps of shell microfabrication (Figure 24A), the filling of shells with drug (Figure 24B), and sealing the shell with a cap (Figure 24C).

Figure 25 is a bar graph showing the stability of IPV (%Recovery) with or without excipients gelatin, maltodextrin, pullulan, myristic acid, and Tween80, during organic/aqueous mixing by vortexing or sonication.

Figure 26 is a bar graph showing the stability of IPV (%Recovery) with or without excipients gelatin, maltodextrin, pullulan, myristic acid, and Tween80, during organic/aqueous mixing with PLGA by vortexing or sonication (son = sonication).

Figure 27 is a bar graph showing the stability of IPV (%Recovery) with excipients pullulan, pullulan/BSA, sorbitol/MSG/MgCl<sub>2</sub>, following drying by overnight lyophilization, or using Genevac for 1 h at 30-35°C.

Figure 28 is a line graph showing change in pH of release medium by PLGA particles over time (days) in the presence of buffering agents Al(OH)<sub>3</sub> (line (2)), myristic acid (line (3)), and Mg(OH)<sub>2</sub> (line (4)). The line representing change in pH of release medium over time in the absence of buffering agents is labeled (1).

Figure 29 is a line graph showing change in pH of release medium by PLGA particles over time (days). The line for PLGA 502H tested after replacing buffer every 2-3 days is designated (1), the line for PLGA 502H tested after replacing buffer every 7 days is designated (2), the line for PLGA 503H tested after replacing buffer every 2-3 days is designated (3), and the line for PLGA 503H tested after replacing buffer every 7 days is designated (4).

## DETAILED DESCRIPTION OF THE INVENTION

### I. Definitions

“Additive manufacturing” or “3D printing” as used herein refers to a process of making a three-dimensional solid object of virtually any shape from a digital model. 3D printing is achieved using an additive process,

where successive layers of material are laid down in different shapes or thicknesses. In some embodiments, “3D printing” involves an extruded or solvent based polymer-containing ink (*e.g.*, PLGA, PLLA, etc.) that is jetted or extruded through a nozzle and solidified into a desired shape. The shape  
5 can be controlled in the x, y and z directions.

“Micromolding,” as used herein, generally refers to processes suitable for manufacturing parts or devices on a microscale, or processes suitable for manufacturing parts or devices having features or tolerances on a microscale. Exemplary techniques include, but are not limited to,  
10 lithography.

“Microdevice,” as used herein, refers to any object or device having microscale dimensions, such as 1 micron to 1000 microns, 1 micron to 500 microns, 1 micron to 250 microns, or 1 micron to 100 microns.

The term “diameter” is art-recognized and is used herein to refer to  
15 either of the physical diameter or the hydrodynamic diameter. The diameter of emulsion typically refers to the hydrodynamic diameter. The diameter of the capsules, both in spherical or non-spherical shape, may refer to the physical diameter in the hydrated state. The diameter of the particles, colloids and cells which are encapsulated inside the capsules refers to the  
20 physical diameter in the hydrated state. As used herein, the diameter of a non-spherical particle or a non-spherical capsule may refer to the largest linear distance between two points on the surface of the particle. When referring to multiple particles or capsules, the diameter of the particles or the capsules typically refers to the average diameter of the particles or the  
25 capsules. Diameter of particles or colloids can be measured using a variety of techniques, including but not limited to the optical or electron microscopy, as well as dynamic light scattering.

The term “biocompatible” as used herein refers to one or more materials that are neither themselves toxic to the host (*e.g.*, an animal or  
30 human), nor degrade (if the material degrades) at a rate that produces monomeric or oligomeric subunits or other byproducts at toxic concentrations in the host.

The term “biodegradable” as used herein means that the materials degrades or breaks down into its component subunits, or digestion, *e.g.*, by a biochemical process, of the material into smaller (*e.g.*, non-polymeric) subunits.

5           The term “microspheres,” “microparticles,” or “microcapsules” is art-recognized, and includes substantially spherical solid or semi-solid structures, *e.g.*, formed from biocompatible polymers such as subject compositions, having a size ranging from about one or greater up to about 1000 microns, such as 1 micron to 500 microns, 1 micron to 250 microns, or  
10   1 micron to 100 microns. The term “microparticles” is also art-recognized, and includes microspheres and microcapsules, as well as structures that may not be readily placed into either of the above two categories, all with dimensions on average of less than about 1000 microns. A microparticle may be spherical or nonspherical and may have any regular or irregular shape. If  
15   the structures are less than about one micron in diameter, then the corresponding art-recognized terms “nanosphere,” “nanocapsule,” and “nanoparticle” may be utilized.

          “Narrow range of release,” as used herein, generally means that the agent is released over a specific period of time, such as an hour, hours, a day,  
20   a week, a month, etc.)

          “Antigen” or “Vaccine,” as used herein, refers to any molecule or entity that produces a specific immune response in a host organism, such as a mammal.

          “Immune response,” as used herein, refers to a specific response to an  
25   antigen or vaccine that produces immunity to any future exposure in a host, such as a mammal.

          “Water soluble” generally refers to something that dissolves or comes apart in aqueous environment.

          “Emulsion” as used herein refers to a liquid discrete phase  
30   homeogeneously dispersed in a liquid continuous phase.

          “Hydrophilic,” as used herein, refers to molecules which have a greater affinity for, and thus solubility in, water as compared to organic solvents. The hydrophilicity of a compound can be quantified by measuring

its partition coefficient between water (or a buffered aqueous solution) and a water-immiscible organic solvent, such as octanol, ethyl acetate, methylene chloride, or methyl tert-butyl ether. If after equilibration a greater concentration of the compound is present in the water than in the organic solvent, then the compound is considered hydrophilic.

“Hydrophobic,” as used herein, refers to molecules which have a greater affinity for, and thus solubility in, organic solvents as compared to water. The hydrophobicity of a compound can be quantified by measuring its partition coefficient between water (or a buffered aqueous solution) and a water-immiscible organic solvent, such as octanol, ethyl acetate, methylene chloride, or methyl tert-butyl ether. If after equilibration a greater concentration of the compound is present in the organic solvent than in the water, then the compound is considered hydrophobic.

## II. Formulations

### A. Polymers and Solvent Systems

#### Polymers

The formulations, which may be formed of microparticles, including microspheres or microcapsules and including those that are emulsion-based, or devices such as those prepared by micromolding, are formed of polymers. Antigen may be dispersed or encapsulated by the polymer. In one embodiment, the device contains a core that only contains one or more vaccines or antigen and stabilizers and the shell or particle wall only contains one or more biodegradable polymers with or without additives. Polymer without antigen may be used to seal or separate areas of the formulation from other areas, and release at different rates.

Polymers must be biocompatible and processible under conditions and using reagents that preserve the antigen. The formulation can be made with hydrophilic polymers, hydrophobic polymers, amphiphilic polymers, or mixtures thereof. The formulation can contain one or more hydrophilic polymers.

Hydrophilic polymers include cellulosic polymers such as starch and polysaccharides; hydrophilic polypeptides; poly(amino acids) such as poly-L-glutamic acid (PGS), gamma-polyglutamic acid, poly-L-aspartic acid,

poly-L-serine, or poly-L-lysine; polyalkylene glycols and polyalkylene oxides such as polyethylene glycol (PEG), polypropylene glycol (PPG), and poly(ethylene oxide) (PEO); poly(oxyethylated polyol); poly(olefinic alcohol); polyvinylpyrrolidone); poly(hydroxyalkylmethacrylamide);  
 5 poly(hydroxyalkylmethacrylate); poly(saccharides); poly(hydroxy acids); poly(vinyl alcohol), and copolymers thereof.

Examples of hydrophobic polymers include polyhydroxyacids such as poly(lactic acid), poly(glycolic acid), and poly(lactic acid-*co*-glycolic acids); polyhydroxyalkanoates such as poly3-hydroxybutyrate or poly4-  
 10 hydroxybutyrate; polycaprolactones; poly(orthoesters); polyanhydrides; poly(phosphazenes); poly(lactide-*co*-caprolactones); polycarbonates such as tyrosine polycarbonates; polyamides (including synthetic and natural polyamides), polypeptides, and poly(amino acids); polyesteramides; polyesters; poly(dioxanones); poly(alkylene alkylates); hydrophobic  
 15 polyethers; polyurethanes; polyetheresters; polyacetals; polycyanoacrylates; polyacrylates; polymethylmethacrylates; polysiloxanes; poly(oxyethylene)/poly(oxypropylene) copolymers; polyketals; polyphosphates; polyhydroxyvalerates; polyalkylene oxalates; polyalkylene succinates; poly(maleic acids), as well as copolymers thereof. In certain  
 20 embodiments, the hydrophobic polymer is an aliphatic polyester. In preferred embodiments, the hydrophobic polymer is poly(lactic acid), poly(glycolic acid), or poly(lactic acid-*co*-glycolic acid).

The formulation can contain one or more biodegradable polymers. Biodegradable polymers can include polymers that are insoluble or sparingly  
 25 soluble in water that are converted chemically or enzymatically in the body into water-soluble materials. Biodegradable polymers can include soluble polymers crosslinked by hydrolyzable cross-linking groups to render the crosslinked polymer insoluble or sparingly soluble in water.

Biodegradable polymers include polyamides, polycarbonates,  
 30 polyalkylenes, polyalkylene glycols, polyalkylene oxides, polyalkylene terephthalates, polyvinyl alcohols, polyvinyl ethers, polyvinyl esters, polyvinyl halides, polyvinylpyrrolidone, polyglycolides, polysiloxanes, polyurethanes and copolymers thereof, alkyl cellulose, hydroxyalkyl



celluloses, cellulose ethers, cellulose esters, nitro celluloses, polymers of acrylic and methacrylic esters, methyl cellulose, ethyl cellulose, hydroxypropyl cellulose, hydroxy-propyl methyl cellulose, hydroxybutyl methyl cellulose, cellulose acetate, cellulose propionate, cellulose acetate butyrate, cellulose acetate phthalate, carboxylethyl cellulose, cellulose triacetate, cellulose sulphate sodium salt, poly (methyl methacrylate), poly(ethylmethacrylate), poly(butylmethacrylate), poly(isobutylmethacrylate), poly(hexylmethacrylate), poly(isodecylmethacrylate), poly(lauryl methacrylate), poly (phenyl methacrylate), poly(methyl acrylate), poly(isopropyl acrylate), poly(isobutyl acrylate), poly(octadecyl acrylate), polyethylene, polypropylene poly(ethylene glycol), poly(ethylene oxide), poly(ethylene terephthalate), poly(vinyl alcohols), poly(vinyl acetate, poly vinyl chloride polystyrene and polyvinylpyrrolidone, derivatives thereof, linear and branched copolymers and block copolymers thereof, and blends thereof. Exemplary biodegradable polymers include polyesters, poly(ortho esters), poly(ethylene imines), poly(caprolactones), poly(hydroxybutyrates), poly(hydroxyvalerates), polyanhydrides, poly(acrylic acids), polyglycolides, poly(urethanes), polycarbonates, polyphosphate esters, polyphosphazenes, derivatives thereof, linear and branched copolymers and block copolymers thereof, and blends thereof.

Amphiphilic polymers can be polymers containing a hydrophobic polymer block and a hydrophilic polymer block. The hydrophobic polymer block can contain one or more of the hydrophobic polymers above or a derivative or copolymer thereof. The hydrophilic polymer block can contain one or more of the hydrophilic polymers above or a derivative or copolymer thereof.

In particularly preferred embodiments the microparticle contains biodegradable polyesters or polyanhydrides such as poly(lactic acid), poly(glycolic acid), and poly(lactic-co-glycolic acid). The microparticles can contain one more of the following polyesters: homopolymers including glycolic acid units, referred to herein as "PGA," and lactic acid units, such as poly-L-lactic acid, poly-D-lactic acid, poly-D,L-lactic acid, poly-L-lactide,

poly-D-lactide, and poly-D,L-lactide, collectively referred to herein as  
“PLA,” and caprolactone units, such as poly( $\epsilon$ -caprolactone), collectively  
referred to herein as “PCL;” and copolymers including lactic acid and  
glycolic acid units, such as various forms of poly(lactic acid-co-glycolic  
5 acid) and poly(lactide-co-glycolide) characterized by the ratio of lactic  
acid:glycolic acid, collectively referred to herein as “PLGA;” and  
polyacrylates, and derivatives thereof. Exemplary polymers also include  
copolymers of polyethylene glycol (PEG) and the aforementioned polyesters,  
such as various forms of PLGA-PEG or PLA-PEG copolymers, collectively  
10 referred to herein as “PEGylated polymers.” In certain embodiments, the  
PEG region can be covalently associated with polymer to yield “PEGylated  
polymers” by a cleavable linker.

The formulation can contain one or a mixture of two or more  
polymers. The microparticles may contain other entities such as stabilizers,  
15 surfactants, or lipids.

### **Solvents**

Solvents must be biocompatible, since some residue will always be  
present in the polymeric formulations. Representative polymer solvents  
include organic solvents such as chloroform, dichloromethane,  
20 tetrafluoroethylene, and acyl acetate. The antigen can be dissolved in  
aqueous or aqueous miscible solvents such as acetone, ethanol, methanol,  
isopropyl alcohol, and mixtures thereof.

### **B. Therapeutic, Prophylactic, Nutraceuticals, and Diagnostic Agents**

25 Although described with reference to delivery of vaccines, it will be  
understood that the formulations may be used to provide release of a variety  
of therapeutic, prophylactic, nutraceutical, and/or diagnostic agents. These  
agents may be low molecular weight drugs, proteins such as hormones or  
growth factors, immunomodifiers, antibodies, nucleic acid molecules (DNA,  
30 RNA, microRNA, siRNA).

## Antigen

### *Infectious Agents*

Antigens for delivery are killed or attenuated infectious agents such as bacteria such as *Clostridia tetani*, viruses such as hepatitis, influenza, and polio, and protozoans such as *Plasmodium* (malaria) and *Leishmania*. Table 2 lists some vaccines the antigens of which can be used in the disclosed formulations. Other antigens are antigenic proteins or haptens such as carbohydrate or sugar antigens effective as antigens for these infectious agents, as cancer antigens, or as immunostimulants.

10 Poliomylitis (Polio) is a highly contagious viral disease that invades the nervous system and can cause total paralysis in a matter of hours. One in 200 infections leads to irreversible paralysis, which is usually confined to the legs. Among those paralyzed, 5% to 10% die due to paralysis of the diaphragm. There is no cure for polio. However, it can be prevented by  
15 vaccination.

Polio cases have decreased by over 99% since 1988, from an estimated 350,000 cases in more than 125 endemic countries to only 223 reported cases in 2012. As of early 2013, only three countries (Afghanistan, Nigeria, and Pakistan) in the world were endemic for the disease. Despite  
20 aggressive vaccination efforts, polio has not been completely eradicated and outbreaks still occur, particularly in developing countries.

There are two types of vaccine that protect against polio: Inactivated Polio Vaccine (IPV) and Oral Polio Vaccine (OPV). To be effective, the IPV needs to be administered to the blood stream. In contrast, the OPV is  
25 effective by crossing intestinal epithelium. While the OPV confers superior intestinal immunity, is easy to administer, and is low in cost, live poliovirus is shed by the vaccinated. This is a concern where the entire population is not vaccinated.

As such, use of an IPV is preferable.

### 30 *Cancer Antigens*

Any protein produced in a tumor cell that has an abnormal structure due to mutation can act as a tumor antigen. Such abnormal proteins are produced due to mutation of the concerned gene. Mutation of

protooncogenes and tumor suppressors which lead to abnormal protein production are the cause of the tumor and thus such abnormal proteins are called tumor-specific antigens. Examples of tumor-specific antigens include the abnormal products of ras and p53 genes. In contrast, mutation of other  
5 genes unrelated to the tumor formation may lead to synthesis of abnormal proteins which are called tumor-associated antigens.

Proteins that are normally produced in very low quantities but whose production is dramatically increased in tumor cells, trigger an immune response. An example of such a protein is the enzyme tyrosinase, which is  
10 required for melanin production. Normally tyrosinase is produced in minute quantities but its levels are very much elevated in melanoma cells.

Oncofetal antigens are another important class of tumor antigens. Examples are alphafetoproteins (AFP) and carcinoembryonic antigen (CEA). These proteins are normally produced in the early stages of embryonic  
15 development and disappear by the time the immune system is fully developed. Thus self-tolerance does not develop against these antigens.

Abnormal proteins are also produced by cells infected with oncoviruses, *e.g.*, Epstein Barr Virus (“EBV”) and Human Papillomavirus (“HPV”). Cells infected by these viruses contain latent viral DNA which is  
20 transcribed and the resulting protein produces an immune response. In addition to proteins, other substances like cell surface glycolipids and glycoproteins may also have an abnormal structure in tumor cells and could thus be targets of the immune system.

Many tumor antigens have the potential to be effective as tumor  
25 vaccines. In addition to alpha fetoprotein (germ cell tumors, hepatocellular carcinoma) and carcinoembryonic antigen (bowel, lung, breast cancers), examples of tumor antigens include CA-125 (ovarian cancer), MUC-1 (breast cancer, epithelial tumor antigen (breast cancer), and melanoma-associated antigen (malignant melanoma).

### 30 C. Stabilizing Agents

Antigen stability is defined as the maintenance of antigen structure during formation of the vaccine formulation and at body temperature. As

discussed below, the polymer composition, selection of solvent, and processing conditions are critical to maintain antigen stability.

Stabilizing agents may also be added. Sugars are a typical group of stabilizing agents for proteins. Examples include simple sugars such as sucrose, fructose, mannitol, glucose, and trehalose as well as more complex sugars. See Alcock *et al.*, Long-term thermostabilization of live poxviral and adenoviral vaccine vectors at supraphysiological temperatures in carbohydrate glass. *Science Translational Medicine*, 2(19):19-19ra12 (2010).

Stabilization of the antigen can be determined by antigen specific ELISA *in vitro* and by measuring the immune response (*e.g.*, IgGs) in animals *in vivo*. Stability is evaluated during each step of the encapsulation and/or manufacturing process, during storage (at 25°C, room temp, high humidity/ high temp conditions, under physiological conditions (pH 7.2, 37°C) and *in vivo* (animal models).

#### **D. Agents increasing rate or completeness of Release**

Gas-generated burst-release systems may allow for instantaneous release of encapsulated antigen. Pore forming agents which are removed by leaching or lyophilization may also be utilized.

### **III. Methods of Manufacture**

It is essential that the methods used to manufacture the device maintain antigen stability, both during processing and at body temperature, and that leakage following formation and administration are minimized. Post-formulation sterilization can typically be accomplished through a combination of sterile manufacturing conditions in combination with methods such as gamma irradiation.

#### **A. Emulsion**

Microparticles can be made using standard techniques. A preferred technique is emulsification of a polymer solution in an organic solvent with an aqueous solution. Addition of organic phase to a large volume of non-solvent phase forms a spontaneous single emulsion and the resulting solution is stirred continuously for solvent evaporation. Immediate formation of microspheres occurs. After stirring, microspheres are washed and then dried.

The examples demonstrate formation of microparticles using emulsification of polymer and antigen, alone or in combination with stabilizers such as trehalose and sucrose.

### **B. Three Dimensional Printing**

5 3D printing could increase consistency of microspheres, allowing for more uniform release, as well as provide a means for making more complex devices such as 'Micro-rods', having increased carrying capacity, that could eliminate the need for simultaneous release from multiple microspheres, as well as facilitate scale up.

10 Three-dimensional (3D) printing is a process of making 3D objects from a digital model. 3D printing is an additive process, where successive layers of material are laid down in different shapes. After each layer is added the "ink" is polymerized, typically by photopolymerization, and the process repeated until a 3D object is created. The recent commercial availability and  
15 reduced cost makes 3D printing of biomolecules, including vaccines and pharmaceuticals, attractive for distribution of these compounds to developing countries. This would negate the need to ship a finished product into the country. Instead, a 3D printer at the point of care can print out the required biomolecule from a simple computer program, which can come from  
20 anywhere in the world.

The 3D printing workflow can be described in 3 sequential steps: 1) the powder supply system platform is lifted and the fabrication platform is lowered one layer; 2) a roller spreads the polymer powder into a thin layer; 3) a print-head prints a liquid binder that bonds the adjacent powder particles  
25 together. Billiet *et al.*, *Biomaterials*, 33:6020-6041 (2012). Two kinds of 3D printing techniques are mostly adopted for nanobiomaterial fabrication. One is inkjet printing with the typical printers. Marizza *et al.*, *Microelectronic Engin.* 111:391-395 (2013). The other is nanoimprint lithography.

Nanoimprint lithography (NIL) is a fast and cost-efficient technique  
30 for fabricating nanostructures. The procedure of NIL is to stack multiple layers of such structures on top of each other; that is, a finished double-layer of structures is covered with a spacer-layer which is planarized using the

chemical-mechanical polishing so that a second layer can be processed on top. Liu *et al.*, *J. Nanomat.* Aug. (2013).

Ink-Jet printing has been used to produce monodisperse PLGA particles. Bohmer *et al.*, *Colloids and Surfaces A: Physiochem. Eng. Aspects*, 5 289: 96-104 (2006). Briefly, droplets of a PLGA solution are printed with the ink-jet nozzle submerged into an aqueous phase. This method produces microspheres at predictable and controllable sizes. This technique has been used to create Paclitaxel-loaded monodisperse microspheres. Radulecu *et al.*, *Digital Fabrication* Sep:18-21. (2005). Variation of this technology has 10 been used to create multilayer monodisperse microspheres. See Kim and Pack, *BioMEMS and Biomedical Nanotechnology*, 1:19-50 (2006). Utilizing this technology, microcapsule shell thickness can be varied from less than 2 microns to tens of microns while maintaining complete and well-centered core encapsulation for microcapsules near 50 microns in overall diameter.

15 Drug delivery rates from microspheres have been varied by providing uniform monodisperse microparticles, mixtures of microparticles of varying sizes, and microparticles having different degradable layers. See Kim and Pack, *BioMEMS and Biomedical Nanotechnology*, 1:19-50 (2006).

Additional information can be found at internet site 20 [store.makerbot.com/replicator2](http://store.makerbot.com/replicator2) (2013); Tekin *et al.*, Inkjet printing as a deposition and patterning tool for polymers and inorganic particles. *Soft Matter*, 4:703-713 (2008); Jang *et al.*, Influence of fluid physical properties on ink-jet printability. *Langmuir*, 25:2629-2635 (2009); Lan, Design and Fabrication of a Modular Multi-Material 3D Printer., M.Sc. Thesis: 25 Massachusetts Institute of Technology, 2013; and at web site [imagexpert.com/site-new/pdf/IXjetXpert.pdf](http://imagexpert.com/site-new/pdf/IXjetXpert.pdf). (2013).

The examples demonstrate preparation of microparticles using 3DP. Waveform was optimized for jetting monodisperse ink droplets consistently. The applied voltage, the duration of the applied voltage, and the change in 30 voltage over time (slope) are all parameters that must be optimized to jet high quality ink drops. For example, the waveform was optimized for a solution of 5% w/v 31k (average molecular weight) PLGA in 1,4-dioxane. Waveforms were optimized with the JetXpert imaging system, and the ink

and waveform and then transferred to the multi-material inkjet 3D-printer. JetXpert imaging was done with a constant pressure waveform. In one embodiment, 5%w/v 31k PLGA/1,4-dioxane ( $Z=5.3$ ,  $\eta=6.08$  mPa-s) was used. The waveform was optimized for this specific solution. In another  
5 embodiment, 15%w/v 12k PLGA/1,2-dichloroethane ( $Z=5.8$ ,  $\eta=6.24$  mPa-s) was used. Most nozzles showed optimal jetting as in the first embodiment, with very few nozzles giving sequences containing satellite drops. In a third embodiment, 15%w/v 12K PLGA/chloroform ( $Z=5.8$ ,  $\eta=5.99$  mPa-s) was used (Table 1). When using more volatile solvents (chloroform or acetone),  
10 most nozzles fired more than one drop continuously. In general, for a constant optimized waveform and similar fluid properties, inks made with more volatile solvents led to inconsistent jetting, and print head nozzles would clog during printing (observed when using 1,2-dichloroethane or chloroform). Inks can be made with 1,4-dioxane and DMF, to prevent  
15 nozzles from clogging. Printing with these solvents requires longer drying time between layers, to prevent structures from morphing. As determined by GPC, for a given polymer, increasing the polymer concentration in the ink increased the ink viscosity. For a given polymer concentration, increasing the molecular weight range of the polymers increased the ink viscosity. After  
20 polymer addition, changes observed in ink densities and surface tensions were negligible relative to viscosity changes.

Figures 11A and 11B are schematics of particles made by emulsion (Figure 11A) and by 3D printing (Figure 11B), and the release of protein from the emulsion particle (Figure 11C) and in an ideal case the 3DP particle  
25 (Figure 11D). These show the differences in the resulting structure, which also changes the release kinetics.

Figures 12A-12D are schematics of the 3D printing process: Creating the structure of PLGA particles (Figures 12A and 24A), filling drugs or proteins into the particles (Figures 12B and 24B), drying the drugs or  
30 proteins (Figure 12C), and encapsulation of the particles (Figures 12D and 24C). The schematic shows a cube shaped structure filled with vaccine, but it is understood that a mold may be utilized to provide any shaped, and that the mold may be filled in layers or more complex patterns.



**Table 1: Solvents for 3DP.**

Mw Range/ Solvent	%w/v	Density (g/ml)	Surface Tension (mN/m)	Viscosity (mPa-s)	Z number (dimension- less)
4k-15k/ Chloroform	5%	1.45	28.07	1.42	24.6
	10%	1.43	27.93	3.17	10.9
	15%	1.44	27.67	5.03	6.9
7k-17k/ Chloroform	5%	1.45	27.99	2.26	15.4
	10%	1.41	27.74	3.64	9.4
	15%	1.43	28.26	5.99	5.8
24k-38k/ Chloroform	5%	1.45	28.12	3.60	9.7
	10%	1.40	27.77	9.32	3.7
	15%	1.44	28.70	26.3	1.3
4k-15k/ 1,2-dichloroethane	5%	1.28	33.31	2.26	15.8
	10%	1.28	32.92	3.65	9.7
	15%	1.27	33.68	5.58	6.4
7k-17k/ 1,2-dichloroethane	5%	1.28	33.69	2.35	15.3
	10%	1.25	32.17	3.88	9.0
	15%	1.29	33.95	6.24	5.8
24k-38k/ 1,2-dichloroethane	5%	1.26	32.36	3.94	8.9
	10%	1.30	33.90	13.8	2.6
	15%	1.27	33.68	49.1	0.7
24k-38k/ 1,4-dioxane	5%	1.09	31.52	6.08	5.3
24k-38k/ dimethylformamide	5%	0.97	36.60	5.50	5.9

Waveform parameters can be optimized to produce uniform single  
5 droplets of “ink” (polymer solution) during printing from a piezoelectric  
nozzle jet (Figure 13). Drop size is a function of various parameters which

are optimized, including applied voltage, slope of voltage, polymer and solvent selection and concentration.

### C. Micromolding

Park *et al.*, *Biomed. Microdevices*, 9:223-234 (2007), describes using  
5 micromolding to fabricate polymer microstructures having sophisticated designs. Micromolds were filled with polymer microparticles, to produce microstructures composed of multiple materials, having complex geometries, and made using mild processing conditions. These microparticles are typically prepared using an oil-water, double-emulsion system; spray drying  
10 methods; supercritical conditioning methods; and milling methods. In a preferred embodiment, micromolds can be prepared by photolithographically creating a female master mold made of photoresist, molding a male master structure out of polydimethylsiloxane (PDMS) from the female master mold and) molding a female replicate mold out of PDMS from the male master  
15 structure. Polymer microparticles can be micromolded using temperature/press methods and/or from solvent.

Polymeric microparticles of 1 to 30  $\mu\text{m}$  in size were made from PLA, PGA and PLGA using spray drying and emulsion techniques. These polymer microparticles were filled into PDMS micromolds at room temperature and  
20 melted or bonded together, for example, by ultrasonically welding microparticles together in the mold while maintaining the voids inherent in their packing structure. Multi-layered microstructures were fabricated to have different compositions of polymers and encapsulated compounds located in different regions of the microstructures. Molds were filled with  
25 solid polymer microparticles instead of a polymer melt to copy microstructures with complex geometries and composed of multiple materials using mild processing conditions. Microparticles can flow easily into the cavities of micromolds at room temperature and low pressure, which facilitates making microstructures with high aspect ratios. Moreover,  
30 polymer microparticles can encapsulate chemical compounds, such as drugs, and can be filled into molds in sequential layers to accommodate multiple material compositions. After filling the mold, the final microstructures can be created by welding the microparticles within the mold by plastic welding

methods, including thermal and ultrasonic welding as well as solvent and gas based welding.

These same techniques can be used to formulate vaccine formulations, once has identified the polymeric materials and conditions required to obtain a narrow time of release at specific time points following administration.

#### IV. Methods of Administration

The vaccine formulations are administered to an individual in need of vaccination. These are administered as a dosage formulation including an effective amount of one or more antigens released in a schedule that elicits a protective effect against the source of the antigen.

Microparticles or microcapsules can be administered by injection, preferably subcutaneously or intramuscularly, for example, under the skin of the back of the upper arm, or to a mucosal surface (orally, intranasally, via the pulmonary route, or other orifice), although injection is preferred if release is to occur over a prolonged period of time of more than a few days.

The dosage form is designed to release a bolus of antigen at the time of administration. This may be achieved by administering a solution or dispersion of antigen in combination with the vaccine formulation providing multiple releases at subsequent times, or the device may be formulated to provide an initial bolus as well as subsequent release(s).

Representative vaccines are shown in Table 2:

Vaccine	Age	Birth	1 month	2 months	4 months	6 months	12 months	15 months	18 months	19-23 months
Hepatitis B <sup>1</sup>		HepB	HepB			HepB				
Rotavirus <sup>2</sup>				RV	RV	RV <sup>2</sup>				
Diphtheria, Tetanus, Pertussis <sup>3</sup>				DTaP	DTaP	DTaP	See footnote <sup>3</sup>	DTaP		
Haemophilus influenza type b <sup>4</sup>				Hib	Hib	Hib <sup>4</sup>	Hib			
Pneumococcal <sup>5</sup>				PCV	PCV	PCV	PCV			
Inactivated Poliovirus <sup>6</sup>				IPV	IPV	IPV				
Influenza <sup>7</sup>						Influenza (Yearly)				
Measles, Mumps, Rubella <sup>8</sup>							MMR		See footnote <sup>8</sup>	
Varicella <sup>9</sup>							Varicella		See footnote <sup>9</sup>	
Hepatitis A <sup>10</sup>							HepA (2 doses)			
Meningococcal <sup>10</sup>										

Additional vaccines of great interest in third world countries include polio and smallpox. The following uses polio vaccine as an exemplary vaccine for this application.

- 5           IPV Vaccine SSI is an inactivated vaccine used for prophylactic vaccination against paralytic poliomyelitis. IPV Vaccine SSI contains inactivated poliovirus type 1, 2 and 3, propagated in Vero cells.

          Contents per dose (0.5ml):

- Inactivated poliovirus type 1 (Brünhilde) 40 D-antigen units;  
10          Inactivated poliovirus type 2 (MEF-1) 8 D-antigen units;  
          Inactivated poliovirus type 3 (Saukett) 32 D-antigen units;  
          Medium 199 to 0.5 ml.

- The vaccine is manufactured without use of serum and trypsin and does not contain preservatives or adjuvants. Antibiotics are not used in the  
15          manufacture. IPV Vaccine SSI contains trace amounts of residual formaldehyde. It is manufactured in Denmark by Statens Serum Institut.

          IPV Vaccine SSI is a solution for injection distributed in single-dose vials. For primary vaccination a series of three doses of 0.5 ml is administered.

- 20          For booster vaccination of previously primary vaccinated persons one dose of 0.5 ml is administered, at the earliest 6 months after the primary vaccination series. Administration of additional booster doses should take place in accordance with national recommendations for polio immunization. The vaccine should be administered intramuscularly or subcutaneously. The  
25          vaccine must not be administered intravascular. The age at the first dose should be at least 6 weeks, and the primary vaccination series should include at least three immunizations, with an interval of at least four weeks. Most countries give IPV using the same schedule as DPT vaccine (typically 2 months, 4 months, and 6 months of age).

- 30          The immunogenicity and safety of IPV Vaccine SSI has been investigated in several clinical trials, including clinical trials with combined vaccines for pediatric use. Apart from IPV these trials included vaccine

antigens against tetanus, diphtheria, pertussis and Haemophilus influenzae type b.

When initiating immunizations at two months of age, completion of a primary vaccination series of three immunizations with at least 1 month  
 5 interval can be expected to result in seroconversion to all three types of poliovirus one month after the second immunization. When initiating immunizations before two months of age, and at the earliest at 6 weeks of age, seroconversion rates between 89% and 99% have been demonstrated. Therefore, in such a schedule, a booster dose at 9 months of age or in the  
 10 second year of life should be considered.

IPV Vaccine SSI can be used for revaccination in infants, pre-school aged children and adults primary immunized with IPV or OPV. IPV Vaccine SSI can be used in mixed IPV/OPV schedules, using one to three doses of  
 15 IPV followed by one to three doses of OPV. It is recommended to administer IPV before the first dose of OPV. In a mixed IPV/OPV schedule, persistence of protective antibodies after primary vaccination has been shown to last at least 20 years. In an IPV only schedule the persistence of protective SSI recommended dose:

20 D-antigen type 1 – 40 DU/ml  
 D-antigen type 2 – 8 DU/ml  
 D-antigen type 3 – 32 DU/ml

10x Trivalent IPV:

25 D-antigen type 1 – 327 DU/ml  
 D-antigen type 2 – 70 DU/ml  
 D-antigen type 3 – 279 DU/ml

The present invention will be further understood by the following non-limiting examples.

### Examples

30 The examples below describe different polymer-drug or polymer-antigen formulations for controlled release of the drug/antigen. The controlled release relies on polymer degradation to achieve multiple bursts of drug/antigen release over time following single injection. Immunogenicity of the formulations following polymer degradation-based bursts of drug/antigen

release and sustained increase in anti-antigen antibody titer are also presented. Methods of making the formulations, which include incorporation of the antigen into the polymer matrix via spontaneous emulsion, or encapsulation of the antigen within a polymer shell via 3D printing or micromolding, are also described. Studies on improving the stability of the antigen, the stability of the polymer matrix, and the stability of the formulations, so that formulations with desired release characteristics and immunogenicity can be obtained, are also presented.

### 10 **Example 1: Selection of Polymers and Solvents for Discrete Release of Vaccine**

#### *Materials and Methods*

Polymer type – PLGA, PLLA; polymer  $M_w$  – 9.5k, 20k, 31k, 46k

Drug loading – 0.5, 3, 5 %; excipients – trehalose, sucrose

Encapsulation – Spontaneous Emulsion

15 The process used  $CH_2Cl_2:TFE::4:1$  ( $CH_2Cl_2$  – dichloromethane TFE – trifluoroethanol) as the organic phase, with poly(vinyl alcohol) (“PVA”) to encapsulate 5%, 3% or 0.5% bovine serum albumin (“BSA”) or inactivated polio virus (“IPV”) into the polymer. The addition of the organic phase to a large volume of a non-solvent phase forms a spontaneous single emulsion and the resulting solution is stirred continuously to evaporate the solvent. Immediate formation of microspheres occurs. After stirring, the microspheres are washed and then dried. See Jaklenec *et al.*, Sequential release of bioactive IGF-I and TGF- $\beta$ 1 from PLGA microsphere-based scaffolds. *Biomaterials*, 29(10):1518-1525 (April 2008).

25 Release studies were done *in vitro* with 10mg of microspheres suspended in 1ml of PBS buffer (pH 7.2) at 37°C. Time points were taken at day 1, 4, 7 and every week thereafter. At each time point, the vials were centrifuged, the supernatant was removed to be assayed, a fresh 1ml of PBS was added and the pelleted microspheres were resuspended.

#### 30 *Results*

Figure 21A is a graph of the release of a model protein, bovine serum albumin (“BSA”) over time (weeks) for 5% BSA, 3% BSA, and 0.5% BSA from PLGA (50:50), 20 kD, derivatized with a carboxylic group.

Figure 21D is a graph of the release of bovine serum albumin over time (weeks) for 5% BSA, 3% BSA, and 0.5% BSA from PLGA (50:50), 9.5 kD, derivatized with a carboxylic group.

5 These figures show release from the same polymer, but with different molecular weights. This difference alone dramatically changed release profiles.

Figures 3A and 3B are graphs of the release of bovine serum albumin over time (weeks) for 5% BSA and 0.5% BSA from PLLA, 50 kD.

10 Figures 4A and 4B are graphs of the release of bovine serum albumin over time (weeks) for 5% BSA and 0.5% BSA from PLLA, 100 kD.

Figures 5A and 5B are graphs of the release of bovine serum albumin over time (weeks) for 5% BSA and 0.5% BSA from PLLA, 300 kD.

15 The only difference between these three graphs is the molecular weight. However, in no case were there substantial, distinct periods of release similar in scope to those in Figure 21.

Figures 6A and 6B are graphs of the release of bovine serum albumin over time (weeks) for 5% BSA and 0.5% BSA from P(d,l)LA, 20 kD. The results are similar to Figures 3-5 and 21.

20 The best results were obtained using PLGA-COOH (50:50). This is summarized in Figures 7A and 7B for PLGA formulations (Figure 7A) compared to bolus injections (Figure 7B).

25 The details of the formulations used for Figure 7A are presented in Table 3 (see also Tables 4 and 5). In Table 3, the difference between F3 and F7 is the Mw of the polymer. F5 has high loading of BSA. Note that F3 and F7 have about 3 equivalent doses, while F5 has a large initial dose followed by two smaller ones—this is very similar to an initial bolus with 2 booster shots. Formulations are injected into animals on day 0 (single injection), while bolus controls are injected 3 times during the course of the study to mimic formulation release kinetics. Negative controls include blank  
30 microspheres for formulations and saline for bolus injections. Injection volumes were 200  $\mu$ l (max volume for subcutaneous (SC) injection in mice). Two injections of 200  $\mu$ l injection, one per leg, were performed. Maximum injectable microsphere concentration was 50 mg/ml.

**Table 3. Amount of BSA released ( $\mu\text{g}$ ) at first, second and third peaks following injection of F3, F5 and F7 formulations.**

Groups (n=10)	Amount of BSA ( $\mu\text{g}$ )			
	1 <sup>st</sup> peak	2 <sup>nd</sup> peak	3 <sup>rd</sup> peak	Total
F3: 0.5% BSA, 20k BI-502H, 50/50	22.2	19.8	21.6	<b>63.6</b>
F7: 0.5% BSA, 31k PLGA, 50/50	23.0	13.6	34.0	<b>70.6</b>
<b>Bolus control (2)</b>	22.0	22.0	22.0	<b>66.0</b>
F5: 5% BSA, 31k PLGA, 50/50	298.0	68.4	65.4	<b>431.8</b>
<b>Bolus control (1)</b>	298.0	68.4	65.4	<b>431.8</b>

5

Figures 8A and 8B are graphs of the release of 0.5% of bovine serum albumin compared to release of 0.5% of ovalbumin over time (weeks) from PLGA (50:50), 20 kD, derivatized with a carboxylic group.

Figures 9A and 9B are graphs of the release of 5% of bovine serum albumin compared to release of 5% of ovalbumin over time (weeks) from PLGA (50:50), 31 kD, derivatized with a carboxylic group.

Figures 10A and 10B are graphs of the release of 0.5% of bovine serum albumin compared to release of 0.5% of ovalbumin over time (weeks) from PLGA (50:50), 31 kD, derivatized with a carboxylic group.

These all shows excellent results, varying only due to the molecular weights of the polymers or the percent loading.

### **Example 2: Predicted Release Profiles from Microparticles made by Emulsion Compared to Microparticles made by 3DP**

Particles fabricated by computer-controlled inkjet 3D-printing can be made with identical micrometer-scale dimensions, drug loadings, and spatial locations of drugs within the polymer microstructures. The main difference between the two methods is that emulsion based particles (left) are matrix based and the drug is homogeneously distributed throughout the particle. The 3D or micromolded particle is shown on the right. The vaccine and polymer are distinctly separated, where the vaccine is in the core and the polymer is only in the shell. These distinctions allow for unique control of release



kinetics between the two particle types. Also, the micromolded/3D printed particle allows for higher loadings since the core size can also be controlled.

The release graph of Figure 11C is based on the data in Example 1.

In the schematic of Figure 11E, the drug or the vaccine is  
5 encapsulated in polymer microspheres and dispersed throughout the polymer matrix (A). Once hydrated, any drug on the surface is immediately released (B) thus causing the initial burst. Following this event, the microspheres have remnant pores through which drug can slowly diffuse causing the secondary burst (C). Finally, when the polymer bulk erodes, due to  
10 significant Mw degradation, the remainder of the drug is released in a third burst (D).

### **Example 3: Selection of Parameters for Piezoelectric Jetting to Make Microparticles**

#### *Materials and Methods*

15 Optimization:

Correct the applied pressures from the nozzles (waveform) to jet out uniform, single droplets of PLGA

Optimize viscosity of the PLGA solution (Z number)

Low enough viscosity to jet out efficiently from the nozzle

20 High enough viscosity to solidify on the substrate

Studies have shown that proper drop formation in piezoelectric drop-on-demand (DOD) ink-jet printing can be described using the dimensionless Z number. The Z number is determined by the ink's fluid properties (surface tension, density, and viscosity) and the size of a printhead's nozzle.

25 Z number ranges can be defined for determining ink printability with a specific waveform.

Characterize PLGA solutions using the Z number:

Solvents are chosen based on PLGA solubility and physical properties (i.e. vapor pressure and boiling point).

30 Once an optimal waveform is determined for one PLGA solution ink, that same waveform can be used to print with different MW PLGAs, by using PLGA solutions that have Z numbers within a certain range.

### *Results*

The waveform for jetting BSA and IPV solutions was optimized as shown in Figure 13. This allows filling of the polymer printer particles with vaccine.

#### 5 **Example 4: Studies to minimize loss of IPV D-antigen and increase stability during lyophilization and encapsulation**

Studies were conducted to optimize methods and reagents for concentrating IPV prior to encapsulation, to prevent loss of D-antigen due to lyophilization. Methods that can be used include centrifugal filters and  
10 dialysis.

Excipients that may reduce D-antigen loss during emulsion process were tested. BSA has been used with some success. The current studies utilized sugars. Varying sugar concentrations for forming sugar glass was tested. Varying humidity conditions for drying was also tested.

15 Comparisons with other solvent systems for the spontaneous emulsion process were also made. In the initial studies, chloroform/acetone was used for the organic phase. In subsequent studies, ethyl acetate was used as a substitute for dichloromethane (DCM).

### *Materials and Methods*

20 Alcock *et al.* reported that viruses incorporated into sugar glass display long-term solid-state stability. (Alcock *et al.*, *Science Translational Medicine*, 2(19):19-19ra12 (2010)) IPV stability in 3D-printed substrates was tested in a similar manner. IPV solutions containing co-dissolved sugars (sucrose and trehalose) were deposited onto a scaled-up PLA structure  
25 printed using the MakerBot Replicator 2. Drying of the IPV/sugar solutions was done at ambient humidity (~20% humidity), which was higher than that used by Alcock *et al.* (~10%).

### *Results*

30 Figures 14A, 14B and 14C are graphs of the percent D-antigen (IPV type I, II, and III) retained after lyophilizing with sugar excipients 1 M trehalose, 1 M sucrose, and 3 M sucrose, then incubating at 4°C, 25°C or 37°C.

The results demonstrate that the sugar significantly increased the stability of the IPV, and that there were few differences between stability at 4°C and 25°C, while stability was less at 37°C. The results also showed that the stability is different depending on the IPV type.

5           Figures 15A and 15B are graphs of the percent D-antigen (IPV type I, II, and III) retained after lyophilizing with 0, 0.5, 0.75, 1 or 1.25 M trehalose, 1 then incubating at 4°C (Figure 15A) or 25°C (Figure 15B). The best results were obtained with 0.5 M and 0.75 M trehalose.

10           Figures 16A and 16B are graphs of the percent D-antigen (IPV type I, II, and III) retained after mixing with solvents until solvent evaporation, with solvents tetrafluoroethylene (“TFE”), dichloromethane (“DCM”), or TFE and DCM, without sugar excipient (Figure 16A) or with 0.75 M trehalose as excipient (Figure 16B). DCM was statistically significantly better. IPV was not stable when exposed to the organic solvents during the encapsulation  
15           process, even when trehalose was used as an excipient.

            Figure 17A is a graph of the percent D-antigen (IPV type I, II, and III) retained after IVP with 0 M sugar, 0.5 M trehalose, or 0.5 M trehalose-sucrose is printed onto 3D-printed PLA substrate and dried at 25°C, 22% RH. Figure 17B is a graph of the percent D-antigen (IPV type I, II, and III)  
20           retained after IVP with 0.25 M trehalose or 0.5 M trehalose-sucrose is printed onto 3D-printed PLA substrate and dried at 25°C, 10.7% RH.

            Alcock *et al.* allowed virus/sugar solution to dry onto substrate under controlled conditions (20-25°C, 2-10% RH). Drying for Figure 17A involved IPV/sugar solutions on PLA substrates within a ventilated cell culture hood  
25           (20-25°C, 22.9% RH). Drying for Figure 17B involved IPV/sugar solutions on PLA substrates in a desiccator (20-25°C, 10.7% RH). Stability was better for room temperature/humidity (all three IPV types had >50% D-antigen retained).

            Figures 18A and 18B are graphs of the percent D-antigen retained  
30           after lyophilization with 0.5 or 0.75 M trehalose then incubating with 25°C (Figure 18A) or after lyophilization with 0.5 M trehalose-sucrose, pipetted, or 0.5 M trehalose-sucrose, jetted, then drying at 25°C, 10.7% RH (Figure 18B).

These studies demonstrate that the best results were obtained by storing the 3D-printed PLA particles containing lyophilized IPV at room temperature and humidity (~20% relative humidity) The studies also demonstrated that jetting the IPV, rather than lyophilizing the IPV, did not significantly decrease IPV stability.

**Example 5: Short-term *in vivo* immunogenicity of BSA-containing PLGA formulations**

The immune response, presented as the anti-BSA IgG (antibody) titer, was measured at 1, 2, and 4 weeks following injection of formulations F3, F5 and F7, presented in Table 3 above and Table 4 below. Formulations F4 and F8 are blank microspheres (no drug), used as negative controls (see Table 4).

The results are presented in Figures 19A, 19B, and 19C. The antibody titer (log<sub>2</sub>) is plotted for the various groups at 1, 2, and 4 weeks. The negative controls are all zero and the microsphere formulations are generating an equivalent or stronger response compared to the bolus control.

**Example 6: Long-term *in vivo* immunogenicity of BSA-containing PLGA microspheres**

*Materials and methods*

*Materials*

Poly(D,L-lactic-co-glycolic acid) (PLGA Resomer® RG 502 H, RG 503 H, RG 504 H, and RG 752 H) and BSA were purchased from Sigma-Aldrich (St. Louis, MO). Poly(vinyl alcohol) (PVA, Mw=25,000) was purchased from Polysciences, Inc. (Warrington, PA). Dichloromethane (DCM) and 2,2,2-trifluoroethanol (TFE) used in this study were reagent grade.

*Microsphere fabrication*

Sixteen formulations of PLGA microspheres containing BSA (Table 4) were fabricated using a spontaneous single-emulsion method (Fu *et al.*, *J Pharm Sci*, 92:1582-1591, 2003; and Jaklenec *et al.*, *Biomaterials*, 29:185-192, 2008). Briefly, 200 mg of PLGA were dissolved in 10 mL of 4:1 DCM:TFE and mixed with 300 µL of BSA in water. Mixing formed a clear, single-phase solution that was then added to 200 mL of 5% (w/v) PVA in

water. The emulsion formed spontaneously and was stirred at room temperature for 3 hours. Particles were then collected via centrifugation, washed five times with water, and lyophilized. When prepared for in vivo use, organic phase and BSA solutions were filtered through 0.2  $\mu\text{m}$  polytetrafluoroethylene filters (Whatman, Little Chalfont, England) and combined in a sterile laminar flow hood.

**Table 4. Microsphere formulations and size characterization.**

Formulation	BSA (% w/w)	PLGA $M_w$ (kDa)	PLGA Ratio	Particle Size ( $\mu\text{m}$ )	90% of Particles Below ( $\mu\text{m}$ )
A (F1)	5	7-17	50:50	10.5 $\pm$ 6.8	18.5
B (F2)	3	7-17	50:50	10.6 $\pm$ 6.4	18.1
C (F3)	0.5	7-17	50:50	10.3 $\pm$ 6.2	22.1
D (F4)	0	7-17	50:50	10.5 $\pm$ 5.9	17.4
E (F5)	5	24-38	50:50	8.6 $\pm$ 6.7	21.4
F (F6)	3	24-38	50:50	11.4 $\pm$ 8.3	21.4
G (F7)	0.5	24-38	50:50	12.1 $\pm$ 8.2	23.1
H (F8)	0	24-38	50:50	11.4 $\pm$ 7.2	20.2
I (F9)	5	38-54	50:50	14.1 $\pm$ 9.4	25.4
J (F10)	3	38-54	50:50	12.0 $\pm$ 7.1	20.3
K (F11)	0.5	38-54	50:50	11.9 $\pm$ 6.8	20.6
L (F12)	0	38-54	50:50	11.9 $\pm$ 6.4	19.7
M (F13)	5	4-15	75:25	11.3 $\pm$ 7.0	19.7
N (F14)	3	4-15	75:25	12.2 $\pm$ 7.4	21.2
O (F15)	0.5	4-15	75:25	12.4 $\pm$ 7.5	21.1
P (F16)	0	4-15	75:25	11.7 $\pm$ 6.5	19.9

10

#### *Microsphere characterization*

Microsphere size distribution was determined using a Multisizer 3 Coulter Counter. Histograms were created using a bin size of 0.39  $\mu\text{m}$  and smoothed using central moving average with a window size of  $\pm 5$  bins.

Scanning electron microscope (SEM) images were collected using a JSM-5600LV SEM (JEOL, Tokyo, Japan) at an acceleration voltage of 5kV. Prior to imaging, samples were coated with Au/Pd using a Hummer 6.2 Sputtering System (Anatech, Battle Creek, MI) to prevent surface charging.

5                                    *In vitro BSA release*

Ten milligrams of microspheres were dispersed into 1 mL phosphate-buffered saline (PBS) in capped tubes and incubated on a rotating platform at 37°C. At each time point (1 day, then weekly for 1-13 weeks), samples were centrifuged at 1500 relative centrifugal force (RCF) for 5 min, after which  
10 the supernatant was collected. Samples were then resuspended in fresh PBS and returned to the incubator for sampling at subsequent time points. BSA release from microspheres was quantified using a bicinchoninic acid (BCA) assay and normalized to the total amount released by the end of the study. Samples were run in triplicate and data reported as mean  $\pm$  standard  
15 deviation.

*In vivo administration of BSA microspheres*

All animal work was approved by MIT's Committee on Animal Care. Briefly, female BALB/c mice between 6 and 8 weeks of age received injections of (1) BSA-loaded microspheres, (2) unloaded microspheres, (3)  
20 bolus BSA, or (4) saline-only. While mice in the first two groups received only one injection, those receiving a bolus BSA or saline-only were injected again at 4 and 8 weeks to match the amount and timing of BSA release from PLGA microspheres in vitro. Samples were dissolved or suspended (when  
25 applicable) in 200  $\mu$ l of saline and injected subcutaneously into each hind limb for a total of 400  $\mu$ l. Table 5 contains the exact dosing regimen for each group. At week 0, 1, 2, 4, 6, 8, and 10, 100  $\mu$ l of blood was sampled sub-mandibularly and, after clotting, was centrifuged at 2000 RCF for 10 minutes at 4°C to separate the serum.

**Table 5. Dosing regimen for *in vivo* BSA administration.**

Group	Amount of BSA ( $\mu$ g)			Total
	First Dose	Second Dose	Third Dose	
Formulation C (0.5% BSA, 7-17 kDa PLGA)*	22	20	22	64
Formulation G (0.5% BSA, 24-38 kDa PLGA)*	23	14	34	71
Low Dose Bolus BSA	22	22	22	66
Formulation E (5% BSA, 24-38 kDa PLGA)*	298	68	65	431
High Dose Bolus BSA	298	68	65	431
Unloaded PLGA microspheres, 7-17 kDa	0	0	0	0
Unloaded PLGA microspheres, 24-38 kDa	0	0	0	0
Saline	0	0	0	0

*All PLGA used in vivo was 50:50. \*Indicates theoretical dose based on in vitro results.*

#### *Immunogenicity of BSA release from PLGA microspheres*

Serum antibody titers against BSA were determined using an endpoint enzyme-linked immunosorbent assay (ELISA). 96-well Maxisorp ELISA plates (Thermo Fisher Scientific, Waltham, MA) were coated overnight at 4°C with 100  $\mu$ L of a 100  $\mu$ g/ml solution of BSA in 0.1M sodium bicarbonate. Plates were then washed three times in PBS containing 0.05% Tween 20 (PBST) and then incubated in 5% non-fat milk in PBST for 2 hours at 37°C as a blocking agent. Following another series of three washes with PBST, mouse serum samples were added in four-fold serial dilutions and incubated for 2 hours at 37°C. The extent of serum dilution increased as the study progressed and titers rose. Plates were then washed five times with PBST and incubated at 37°C with horseradish peroxidase-conjugated goat

anti-mouse secondary antibody (Southern Biotechnology Associates, Birmingham, Alabama) diluted 1:1000 in blocking buffer for 2 hours. Plates were washed an additional five times with PBST and developed using 100  $\mu$ L of p-nitrophenyl phosphate solution prepared from tablets dissolved in 1x diethanolamine buffer from an alkaline phosphate substrate kit (Bio-Rad, Hercules, CA). After 10 minutes, the reaction was stopped by adding 100  $\mu$ L of 0.4M sodium hydroxide to each well and absorbance values were read at 405 nm using a Tecan Infinite M200 Pro microplate reader (Männedorf, Switzerland). Titers were reported as the reciprocal of the highest serum dilution that yielded an absorbance greater than 2-fold above background values.

#### *Statistical analysis*

All data in Table 4 and Figure 20 are reported as mean  $\pm$  standard deviation. In vitro studies were performed in experimental triplicate while in vivo studies were performed with ten experimental replicates for all groups at all time points, except Formulation C at week 10 in which nine replicates were used due to insufficient blood volume from one animal. Time-matched antibody titers were compared using one-way ANOVA statistical analysis for the three low dose groups and Student's unpaired two-tailed t-test for comparing the high dose microspheres and bolus injections. Comparisons of peak antibody titers were performed using Student's unpaired two-tailed t-test with the Holm-Bonferroni correction at a significance level of 0.05 to counteract the effect of multiple comparisons.

#### *Results*

##### *Characterization of BSA-containing PLGA microspheres*

All sixteen formulations of PLGA and BSA produced spherical microparticles with broad distributions of particle sizes (Table 4). The size and shape of Formulations C, G, and E, which were used for in vivo studies, were representative of all formulations. Formulation C produced microspheres that were  $10.3 \pm 6.2 \mu\text{m}$  in diameter yet because of the cubic relationship between volume and diameter, particles smaller than  $10.3 \mu\text{m}$  contained just 4.2% of the antigen load assuming homogeneous distribution of BSA in PLGA (Narasimhan *et al.*, *J Control Release*, 47:13-20, 1997).



Larger particles contained a majority of the antigen with 90% of the volume contained in particles larger than 22.1  $\mu\text{m}$  for Formulation C. Formulations G and E demonstrated similar characteristics with particle diameters of  $12.1 \pm 8.2$  and  $8.6 \pm 6.7$   $\mu\text{m}$  respectively, yet with 90% of particle volume contained in particles larger than 23.1 and 21.4  $\mu\text{m}$  respectively. Histograms of microsphere diameter and volume distribution can be seen in Figure 20A-20F. Particles at the large end of the distribution also contributed substantially to surface area effects as 50% of the cumulative particle surface area was present on particles larger than 23.67, 29.54, and 27.58  $\mu\text{m}$  in diameter for Formulations C, G, and E respectively.

#### *In vitro release of BSA from PLGA microspheres*

*In vitro* release kinetics from PLGA microspheres were determined by BCA assay and expressed as a percentage of the total BSA released during the duration of the experiment. BSA release and particle degradation occurred more quickly in PLGA microspheres fabricated using any of the three polymers with 50:50 ratios of lactic-to-glycolic acid compared to a 75:25 ratio. All microsphere formulations produced using PLGA with a 50:50 ratio degraded within 14 weeks whereas 75:25 PLGA degraded in 22 weeks. The timing of BSA release appeared to be more dependent on polymer type than BSA loading, though loading had a major effect on the size of the bursts in most cases. Low molecular weight (7-17 kDa) PLGA released in three distinct bursts over the course of 8 weeks and completely degraded by week 14 as seen in Figure 21A. The medium molecular weight (24-38 kDa) PLGA also displayed three bursts spread over 9-12 weeks depending on BSA loading and degraded by week 14 (Figure 21B). The highest molecular weight (38-54 kDa) PLGA released BSA over 9-12 weeks with prominent bursts at day 1 and week 8, but with more continuous release kinetics in between from the 3 and 5% BSA loaded microspheres (Figure 21C). Microspheres made from low molecular weight (4-15 kDa) PLGA with a higher lactic acid content (75:25) degraded over 22 weeks and showed gradual semi-continuous release after an initial burst (Figure 21D).

Out of the sixteen *in vitro* formulations, Formulations C, G, and E were chosen for the subsequent *in vivo* study based on their *in vitro* release

kinetics. Formulation C microspheres were made using low molecular weight PLGA (7-17 kDa) loaded with 0.5% BSA, Formulation G with slightly higher molecular weight PLGA (24-38 kDa) and 0.5% BSA, and Formulation E with the same molecular weight PLGA as Formulation G, but with higher (5%) BSA loading. BSA release from Formulation C was characterized by three distinct peaks at day 1 when  $33.4 \pm 5.1\%$  of total BSA was released, at week 4 when  $27.2 \pm 4.3\%$  was released, and across the week 6 and week 7 time points during which  $32.1 \pm 5.2\%$  was released respectively (Figure 21A). Minimal BSA release was observed at weeks 1, 2, 4, and 5, or after week 7 and microspheres were completely degraded by week 10 as evidenced by complete dissolution of the particles.

Formulation G microspheres were also characterized by BSA release in three bursts (Figure 21B), but spread out over a longer timeframe. In addition, total microsphere degradation was observed after 14 weeks rather than 10 weeks for Formulation C. The first BSA burst from Formulation G was observed at the 1-day time-point, releasing  $28.0 \pm 5.7\%$  of total BSA. This was followed by a second burst of  $12.9 \pm 2.9\%$  at week 4, and an elongated burst at weeks 8 through 11 during which a cumulative  $43.8 \pm 2.0\%$  of BSA was released. Little-to-no BSA was observed in the release media at any other time points through complete degradation of the particles.

Formulation E microspheres released  $63.7 \pm 7.3\%$  of its BSA at day 1 which was the largest initial burst in terms of both total quantity and percentage of any of the formulations (Figure 21C). The only time points with substantial BSA release following this initial burst were weeks 3 and 8 when  $8.9 \pm 0.9\%$  and  $8.8 \pm 2.1\%$  of the total load was released respectively. Similar to Formulation G, which used PLGA at the same molecular weight (24-38 kDa), Formulation E degraded completely at 14 weeks.

#### *Immunogenicity of BSA-containing PLGA microspheres*

The humoral immune response to each microsphere formulation was compared with a positive control consisting of three bolus injections matching the quantity and timing of BSA released from particles *in vitro*. Within experimental groups, a statistically significant increase in titer between consecutive weeks was used as evidence of release (one-sample

paired t-test). Formulation C induced a significant increase in antibody titer compared to the previous time point at weeks 1, 2, and 4 ( $p < 0.05$ ,  $p < 0.0001$ , and  $p < 0.001$  respectively), then decreased significantly at weeks 6 and 8 ( $p < 0.01$  and  $p < 0.05$  respectively), before stabilizing at week 10 (Figure 22).

5 Similarly, mice receiving Formulation G showed a significant increase in titer at weeks 1, 2, and 4 ( $p < 0.05$ ,  $p < 0.001$ , and  $p < 0.01$  respectively), remained steady at week 6, and then fell significantly by week 8 ( $p < 0.05$ ) before stabilizing again at week 10. Formulation E induced a similar response as antibody titers increased significantly at weeks 1, 2, and 4

10 ( $p < 0.001$  for all), leveled off at week 6, and then decreased through the end of the study ( $p < 0.05$ ). Overall, the immune response to all three microparticle formulations demonstrated a similar progression over time as titers rose over the first 4 weeks then slowly decreased through week 10 (Figure 22). However, the magnitude of antibody titers appeared highly

15 dependent on BSA loading. Based on in vitro experiments, Formulation E released approximately 13 times more BSA than Formulations C at the earliest time point (1 day) due to a large initial burst and induced antibody titers that was 13-fold higher as well. This trend was also observed at the end of the study as the antibody titer induced by Formulation E was 8-fold higher

20 than those associated with Formulation C after releasing 7 times as much BSA.

Antibody titers from animals receiving any of the BSA-loaded microsphere formulations were significantly higher than those from animals receiving dose-matched bolus injections after 2 and 4 weeks ( $p < 0.05$  and

25  $p < 0.01$  respectively). However, after boosting with a second bolus injection at week 4, titers were statistically similar between formulations and their dose-matched bolus group at 6 and 8 weeks. Then at week 10, following administration of the third bolus, animals receiving bolus BSA showed another spike in titer resulting in significantly higher antibody titers

30 compared to all dose-matched microsphere groups at that time point ( $p < 0.001$ ).

While time-matched antibody titers suggest the timing of antigen presentation, peak antibody titers may be more useful indicators of immune

response in this case due to a possible mismatch in the timing of antigen presentation (Paryani *et al.*, *J. Pediatr.*, 105:200-205 (1984); Barraclough *et al.*, *Am J Kidney Dis.*, 54:95-103 (2009)). The bolus injection schedule was chosen to match the timing of microsphere bursts in vitro, but accelerated in vivo degradation could have resulted in an offset in antigen release. As a result, at the end of the experiment antibody titers in the microsphere groups had been falling for weeks as would be expected in the absence of antigen, whereas titers in the bolus control groups reached their highest levels following the third injection. Comparing peak antibody titers helps to control for the effects of timing and determine which treatment was more immunogenic.

Antibody titers for Formulations C, G, and E peaked at  $13.9 \pm 1.3$ ,  $13.7 \pm 2.2$ , and  $16.1 \pm 2.1$  on a log<sub>2</sub> scale respectively 4 weeks after microsphere administration, whereas the small and large dose-matched boluses peaked after 10 weeks at  $15.5 \pm 1.5$  and  $17.7 \pm 0.8$  log<sub>2</sub> titer respectively (Figure 23). Formulations C and G (Figure 23A) induced peak antibody titers that were not statistically different ( $p^*=0.0645$  and  $p^*=0.0543$  respectively) than the dose-matched bolus control consisting of three 22  $\mu$ g BSA injections, using the Holm-Bonferroni correction method that has been recommended for multi-group titer comparisons (Reverberi, *Blood Transfus.* 6:37-45 (2008)). Formulation E (Figure 23B) also induced peak titers that were not statistically different ( $p^*=0.0784$ ) than the dose-matched bolus control consisting of three bolus injections (Table 5).

#### **Example 7: PLGA micromolded particles for drug delivery**

##### **Materials and Methods**

Development of single drug particles involves three major steps: 1) shell microfabrication, 2) vaccine/drug filling and 3) particle sealing.

The shell microfabrication may use mask fabrication (lithography), fluoro-mold fabrication (UV core) and heat pressing of PLGA (120°C for 30 min). The step of drug filling may be carried out by the Biodot robot, filling the core with 1.5 nL volume of the drug. The sealing is achieved by placing a PLGA cap on the shell and then sealing the particles at 37°C for 5 min and acetone vapor for 5 min.

The schematic of the process is presented in Figures 12A-12D and 24A-24C. These microparticles may be manufactured as stacks, sealing each one with another at 37°C for 5 min.

### ***Results***

5 The micromolded particle shells have an x, y, and z dimensions of 450 x 450 x 300  $\mu\text{m}$  with a core of 100 x 100 x 100  $\mu\text{m}$  (cap dimensions are 450 x 450 x 150  $\mu\text{m}$ ), or 200 x 200 x 150  $\mu\text{m}$ , with a core of 100 x 100 x 100  $\mu\text{m}$ . These dimensions allow the particles to be delivered with a 21 gauge  
10 needle (inner diameter 514  $\mu\text{m}$ ) or with a 23 gauge needle (inner diameter 337  $\mu\text{m}$ ).

### **Example 8. Single PLA particles for depositing drug/sugar solution**

#### ***Materials and methods***

IPV in 0.5 M trehalose and 0.5 M sucrose solution was deposited onto PLA particles then dried in the stem cell culture hood at room  
15 temperature. PLA particles may be 3D-printed using a Makerbot printer.

### ***Results***

Single PLA particles may serve as a useful platform for drying onto and delivering drugs and antigens, as this method allows to deposit drug or antigen in a sugar solution onto a substrate such that the drug or antigen is  
20 stabilized within a sugar glass. Studies on the stability of IPV antigens on PLA particles are presented in Examples below.

### **Example 9: Stability of the IPV microspheres**

The process of IPV microsphere formation presented in Example 1 above holds challenges for IPV stability during the manufacturing process.  
25 These challenges with a summary of the findings that overcome the challenges are presented in Table 6.

**Table 6. Summary of obstacles to IPV stability, approaches taken to overcome the obstacles and findings.**

Obstacle	Approach	Findings
IPV is not stable in organic solvents or at organic/aqueous interface	Double emulsion method keeps IPV in an aqueous "bubble." Minimal direct contact with organics	<ul style="list-style-type: none"> <li>- Double emulsion is effective</li> <li>- Vortexing and sonication parameters can be tuned to maximize IPV recovery</li> <li>- Excipients improve recovery               <ul style="list-style-type: none"> <li>- Gelatin</li> <li>- Tween (polysorbate) 80</li> </ul> </li> </ul>
	Co-encapsulate IPV with BSA or mild surfactant to reduce IPV exposure to aqueous/organic interface during emulsification	
IPV may not survive the physical mixing processes that form the emulsion	Use lower-energy mixing: homogenization, vortexing, low-amplitude sonication, etc.	
IPV is not stable after drying	Change drying method (lyophilization, air-drying, vacuum)	- Microspheres dried at room temp for 1 hr under vacuum show high IPV recovery
	Add excipients during drying	- Sorbitol/MSG/MgCl <sub>2</sub> improves recovery during drying
	Co-encapsulate with excipients	- Sorbitol/MSG/MgCl <sub>2</sub> can be co-encapsulated in w/o/o microspheres
IPV is not stable in microsphere matrix in hydrated, 37C environment	Co-encapsulate IPV with excipients to help maintain stability	- Gelatin and sorbitol/MSG/MgCl <sub>2</sub> improve stability over time
IPV is not stable in polymer degradation products	Co-encapsulate IPV with basic excipients to buffer acidic byproducts	<ul style="list-style-type: none"> <li>- IPV is not stable &lt;pH 6 or &gt;pH 8</li> <li>- Mg(OH)<sub>2</sub> buffers acidic byproducts</li> </ul>
Microspheres do not show bursts of release at the desired time points	Change polymer composition (PLGA MW, ratio of LA:GA)	- PLGAs of different MW and ratios are being tested with BSA microspheres
	Change excipients added to polymer	- Microspheres containing pH-affecting additives are being tested

**5 Example 10: Co-encapsulation of IPV with gelatin or mild surfactant and sonication increase IPV stability during emulsification**

*Materials and Methods*

IPV was mixed with water and/or excipients at 100:1 w/w ratio. Excipients used were gelatin, maltodextrin, pullulan, myristic acid, and Tween80. IPV with excipients was then added to DCM, or PLGA/DCM. The mixture was vortexed for 10 sec on highest setting, or was sonicated 10 sec at 25% amplitude. ELISA buffer (1% BSA, 1% Triton-X 100 in PBS) was added, the mixture was vortexed for 10 sec on high setting, and layers were

separated by centrifugation. The aqueous layer was removed and tested by ELISA.

### ***Results***

Gelatin and Tween80 both improved the stability of type I, II and III  
5 IPV (Figure 25).

Type I and III IPV, generally the most sensitive to damage, showed best recovery with sonication at 20% amplitude for 30 sec (Figure 26).

### **Example 11: Drying of IPV and excipients with Genevac improves IPV recovery**

#### 10 ***Materials and methods***

IPV was concentrated 26x and mixed with pullulan, pullulan/BSA, or sorbitol/MSG/MgCl<sub>2</sub> at 100:1, 350:1, or 700:1 mass ratio. IPV with excipient was dried by lyophilization (overnight) or using Genevac (1 hr, 30-35°C). Recovery was tested by ELISA after resuspending the dried IPV with  
15 excipients.

#### ***Results***

In all paired groups of lyophilization (Lyo) and Genevac (Figure 27), Genevac showed higher recovery than lyophilization.

### **Example 12: Lyophilized IPV shows long-term stability at 37°C**

#### 20 ***Materials and methods***

IPV was mixed with excipients 10% sorbitol, 8.5% MSG, and 8.5% MgCl<sub>2</sub> at 40,000:1, and lyophilized in plastic tube. The lyophilized IPV with excipients was stored in a plastic tube in a pouch with desiccant at humidified atmosphere and 37°C for up to 30 days. At various time points,  
25 the lyophilized IPV was resuspended and tested for recovery by ELISA.

#### ***Results***

Types I, II and III IPV in lyophilized form with excipients showed long-term stability when stored at 37°C (Figure 2).

### **Example 13: Changes in pH of release medium by PLGA particles**

30 PLGA degrades by bulk erosion leading to acid buildup inside microspheres. During *in vitro* storage, finite volume of buffer may cause acid build up in tubes, which may lead to particle erosion and acid build up inside the microspheres. In *in vivo* applications, the buffer in local environment is

constantly replenished with minimal acid build up outside of the particles. Because PLGA degrades by bulk erosion, acid build up inside the microspheres may be a problem. Therefore, several PLGA molecules were tested for changes in the pH of the release medium over time, when the  
5 buffer was changed every 2-3 days, or every 7 days.

### ***Results***

Results in Figure 29 demonstrate that PLGA molecules alone change the pH of the release medium making it more acidic over time. Although change of buffer every 2-3 days helps ameliorates this effect, it does not  
10 solve the problem.

### **Example 14. Excipients to buffer the acidic products of PLGA microspheres**

It was observed that Type I IPV was relatively stable at pH 7.4 at 37°C, and type II and type III IPV were stable at pH 6-8 at 37°C. Therefore,  
15 several excipients were tested for their buffering capacity to buffer the acidic products of PLGA microspheres.

### ***Materials and Methods***

Buffering agents, such as  $Mg(OH)_2$ ,  $Al(OH)_3$ , and myristic acid, were incorporated into PLGA microspheres to minimize changes in pH of release  
20 medium. Also,  $Mg(OH)_2$  has little to no solubility in water, it remains distributed throughout the polymer matrix and can potentially buffer the internal compartment.  $Al(OH)_3$  is a known adjuvant and can increase immunogenicity

### ***Results***

The results in Figure 28 demonstrate that  $Mg(OH)_2$  was able to buffer  
25 acidic byproducts in the release medium and stabilize the pH of the release medium at about pH 7.

### **Example 15. IPV microspheres incorporating stabilizing excipients**

Based on the investigations of Examples 9-14, PLGA microspheres  
30 incorporating IPV and various stabilizing additives were generated. These formulations are presented in Table 7 below, and were made with PLGA 502H. Other formulations with higher MW polymers, like PLGA 503H, are also contemplated.



**Table 7. IPV microspheres formulations incorporating stabilizing additives for polymer degradation-based bursts of drug/antigen release. S/M/M refers to 10% sorbitol, 8.5% MSG, 8.5% MgCl<sub>2</sub>.**

mg polymer (MW)	Approx. DU IPV loaded (Type I/Type II/Type III)	Additives to IPV phase	Additives to PLGA phase	Outer stirring phase (2 <sup>nd</sup> emulsion)	Mixing (1 <sup>st</sup> emulsion)	Mixing speed/time	Drying
20 (12k)	216/62/187	4 mg gelatin	None	Oil: mineral oil + Span 80	Sonicator	20% amplitude, 30 sec	22 mg S/M/M
20 (12k)	216/62/187	4 mg gelatin	10% myristic acid	Aqueous: 1% PV A, 5% sucrose	Sonicator	20% amplitude, 30 sec	22 mg S/M/M
20 (12k)	216/62/187	4 mg gelatin	10% Al(OH) <sub>3</sub>	Aqueous: 1% PV A, 5% sucrose	Sonicator	20% amplitude, 30 sec	22 mg S/M/M
20 (12k)	216/62/187	8.5 mg S/M/M	None	Oil: mineral oil + Span 80	Sonicator	20% amplitude, 30 sec	No excipient
20 (12k)	216/62/187	4 mg gelatin	10% Mg(OH) <sub>2</sub>	Aqueous: 1% PV A, 5% sucrose	Sonicator	20% amplitude, 30 sec	22 mg S/M/M
20 (12k)	216/62/187	2 mg gelatin, 4.3 mg S/M/M	10% Mg(OH) <sub>2</sub>	Oil: mineral oil + Span 80	Sonicator	20% amplitude, 30 sec	No excipient
20 (12k)	216/62/187	4 mg gelatin	10% Mg(OH) <sub>2</sub>	Oil: mineral oil + Span 80	Sonicator	20% amplitude, 30 sec	No excipient
20 (31k)	216/62/187	4 mg gelatin	10% Mg(OH) <sub>2</sub>	Oil: mineral oil + Span 80	Sonicator	20% amplitude, 30 sec	No excipient
20 (31k)	216/62/187	2 mg gelatin, 4.3 mg S/M/M	10% Mg(OH) <sub>2</sub>	Oil: mineral oil + Span 80	Sonicator	20% amplitude, 30 sec	No excipient
20 (12k)	216/62/187	4 mg gelatin	10% Mg(OH) <sub>2</sub> , 10% myristic acid	Oil: mineral oil + Span 80	Sonicator	20% amplitude, 30 sec	No excipient
20 (12k)	216/62/187	4 mg gelatin	5% Mg(OH) <sub>2</sub> , 5% Al(OH) <sub>3</sub>	Oil: mineral oil + Span 80	Sonicator	20% amplitude, 30 sec	No excipient

**Example 16. IPV stability after drying on PLA allows use of lower excipient:vaccine ratios**

D-antigen retention after drying 26X Trivalent IPV on PLA for 16 hours at room temperature and humidity was tested. Excipient used was 10% sorbitol, 8.5% MSG, 8.5% MgCl<sub>2</sub>. The ratios of excipient to vaccine tested were 300 to 1, 100 to 1, 60 to 1, 30 to 1, or no excipient. The percent of D-antigen retained was assayed by ELISA.

**Results**

Upon drying of the solutions on PLA, no significant drop in stability was observed when going down to a 30:1 ratio (Figure 1). This data indicated that concentrated excipient-free IPV could be dried on PLA cubes without much loss of antigen activity.

**Example 17. Other formulations of concentrated IPV mixed with excipients and dried in a PLA cube**

The formulations listed in Table 8 represent % D-antigen retained when concentrated IPV was mixed with indicated excipient solutions (columns "Polyol" and "Other") at the indicated ratios (column "Excip:Vax," all indicated ratios are 100:1) and dried in PLA cubes overnight.

Unless defined otherwise, all technical and scientific terms used herein have the same meanings as commonly understood by one of skill in the art to which the disclosed invention belongs. Publications cited herein and the materials for which they are cited are specifically incorporated by reference. Those skilled in the art will recognize, or be able to ascertain using no more than routine experimentation, many equivalents to the specific embodiments of the invention described herein. Such equivalents are intended to be encompassed by the following claims.

**Table 8. Percent D-antigen (type 1, type 3 and type 3 IPV) retained on PL.A particles after air-drying for one day with the indicated excipients and IPV at a ratio of 100:1.**

Polyol	Other	Excip:Vax	Days	°C	Drying	Storage	Type 1	Type 2	Type 3
none	Silk	100	1	25	Air-dry	BSL cabinet	4	18	4
Maltodextrin	Arginine	100	1	25	Air-dry	BSL cabinet	7	84	38
Methyl Cellulose	Ectoin	100	1	25	Air-dry	BSL cabinet	62	52	57
HPMC	Ubiquitin	100	1	25	Air-dry	BSL cabinet	59	n/a	79
HPMC	MSG	100	1	25	Air-dry	BSL cabinet	65	n/a	73
none	Gelatin, ubiquitin	100	1	25	Air-dry	BSL cabinet	75	74	79
Sucrose	Gelatin	100	1	25	Air-dry	BSL cabinet	77	78	79
Ca Heptagluconate	k Phosphate	100	1	25	Air-dry	BSL cabinet	78	80	86
Trehalose	Threonine	100	1	25	Air-dry	BSL cabinet	81	83	84
CMC	Gelatin	100	1	25	Air-dry	BSL cabinet	82	82	84
Glycerol	MSG	100	1	25	Air-dry	BSL cabinet	82	84	86
Ca-Heptagluconate	Gelatin	100	1	25	Air-dry	BSL cabinet	85	n/a	83
Sorbitol	Peptone	100	1	25	Air-dry	BSL cabinet	84	87	85
Trehalose	Ubiquitin	100	1	25	Air-dry	BSL cabinet	84	91	90
Mannitol	Glutamine	100	1	25	Air-dry	BSL cabinet	91	n/a	89
Sorbitol	Glycine	100	1	25	Air-dry	BSL cabinet	90	88	90
Maltodextrin	Glutamine	100	1	25	Air-dry	BSL cabinet	88	n/a	94
Maltodextrin	none	100	1	25	Air-dry	BSL cabinet	93	95	91
Mannitol	Ectoin	100	1	25	Air-dry	BSL cabinet	92	94	92
Trehalose	Glycine	100	1	25	Air-dry	BSL cabinet	93	90	92
Sorbitol	Ectoin	100	1	25	Air-dry	BSL cabinet	96	n/a	92
Sucrose	Threonine	100	1	25	Air-dry	BSL cabinet	95	n/a	97
Sorbitol	MSG, MgCl2	100	1	25	Air-dry	BSL cabinet	99	95	99

## CLAIMS

We claim:

1. A polymeric formulation formed by emulsion formulation, three dimensional printing, or micromolding of a biocompatible polymer comprising antigen, optionally in combination with stabilizer for the antigen, wherein the antigen is released in a defined time period in an effective amount to elicit an immune response *in vivo*.
2. A polymeric formulation formed by emulsion formulation, three dimensional printing, or micromolding of a biocompatible polymer comprising therapeutic, prophylactic, nutraceutical, or diagnostic agent, optionally in combination with stabilizer, wherein the agent is released from discrete regions of the formulation at at least two different time points or with different release kinetics.
3. The formulation of claim 2, wherein the agent is one or more antigens, wherein the antigen is released in a defined time period in an effective amount to elicit an immune response *in vivo*.
4. The formulation of claim 1 or 3 further comprising an effective amount of antigen to elicit an immune response *in vivo* which is present in excipient, on the surface or, or mixed with, the formulation at the time of administration.
5. The formulation of claim 1 or 2 further comprising a stabilizing agent selected from the group consisting of sugars, oils, lipids, and carbohydrates, preferably sugar, most preferably a sugar selected from the group consisting of sucrose, trehalose, and combinations thereof, optionally in the form of a sugar glass.
6. The formulation of claim 1 or 3, wherein the antigen elicits an immune response to an infectious agent or to a tumor.
7. The formulation of claim 6, wherein the infectious agent is a virus, bacteria, fungus or protozoan.
8. The formulation of claim 7, wherein the virus is selected from the group consisting of polio, influenza, hepatitis, rotavirus, measles, mumps, rubella, and varicella.

9. The formulation of claim 7, wherein the bacteria is selected from the group consisting of Diphtheria, Pertussi, Clostridium tetani, Pneumococcus, and Meningococcus.
10. The formulation of claim 6, wherein the antigen is a tumor antigen selectively eliciting a T cell response to a tumor.
11. The formulation of claim 1 or 2, wherein the polymer is biodegradable by hydrolysis, preferably a polyester, most preferably selected from the group consisting of poly(lactic acid), poly(glycolic acid), and copolymers thereof.
12. The formulation of claim 1 or 2 providing release at intervals of ten to ninety days, preferably at intervals of 30 to 60 days.
13. The formulation of claim 1 or 2 which is in the form of microparticles, microcapsules, or microspheres.
14. The formulation of claim 1 or 2, wherein the formulation is injectable.
15. The formulation of claim 1 or 2, wherein the formulation is implantable.
16. The formulation of claim 1 or 2, wherein the formulation can be applied to a mucosal surface selected from the group consisting of nasal, pulmonary, oral, vaginal and rectal.
17. A method of delivery a therapeutic, prophylactic, nutraceutical or diagnostic agent to an individual in need thereof, comprising administering to the individual the formulation of any of claims 1-16.
18. The method of claim 17, wherein the formulation is a polymeric formulation formed by three dimensional printing or micromolding of a biocompatible polymer comprising antigen, optionally in combination with stabilizer for the antigen, wherein the antigen is released in a defined time period in an effective amount to elicit an immune response *in vivo*.
19. The method of claim 18, wherein the antigen is released at two or more monthly intervals, and the antigen is polio or DPT.

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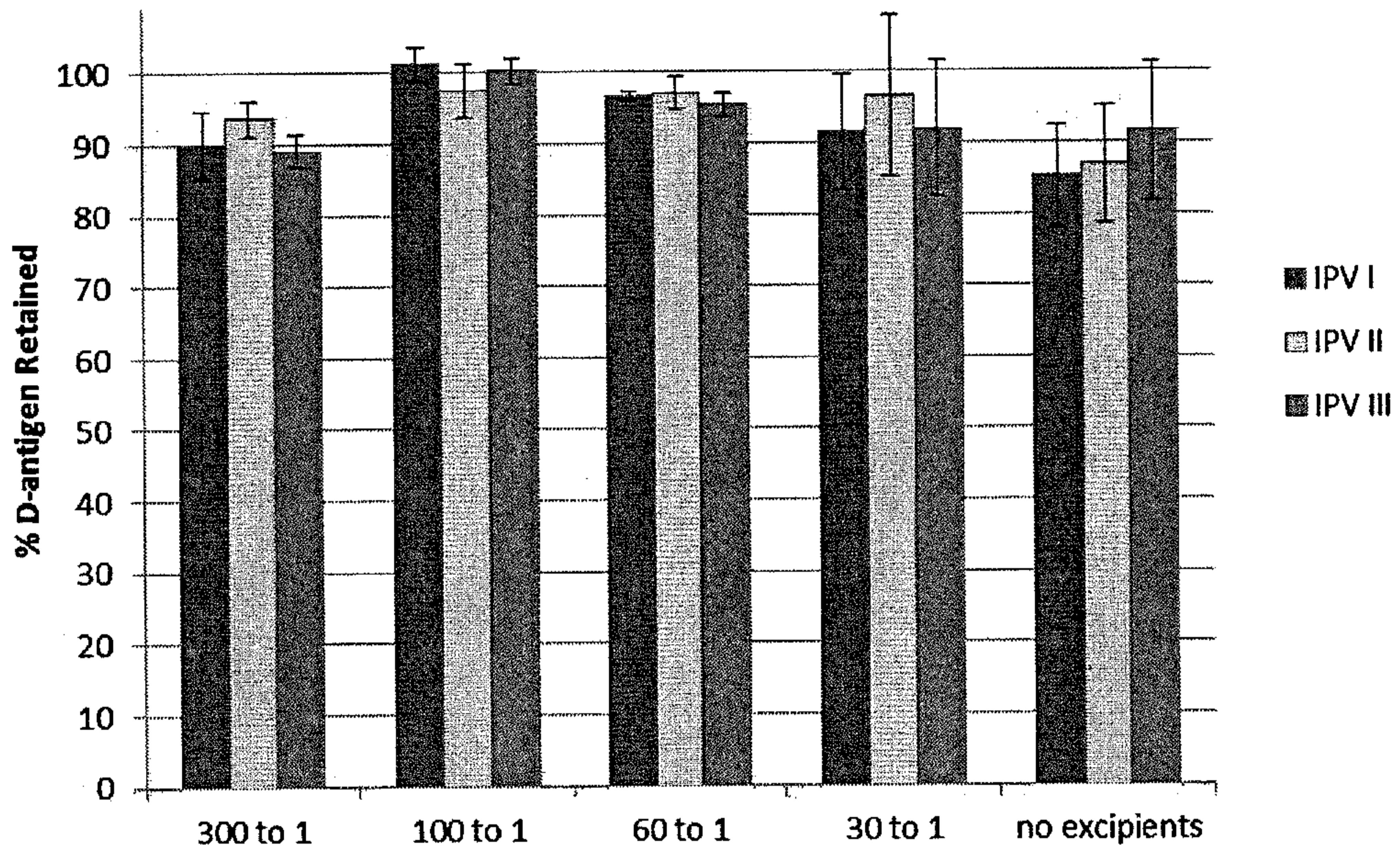


FIG. 1

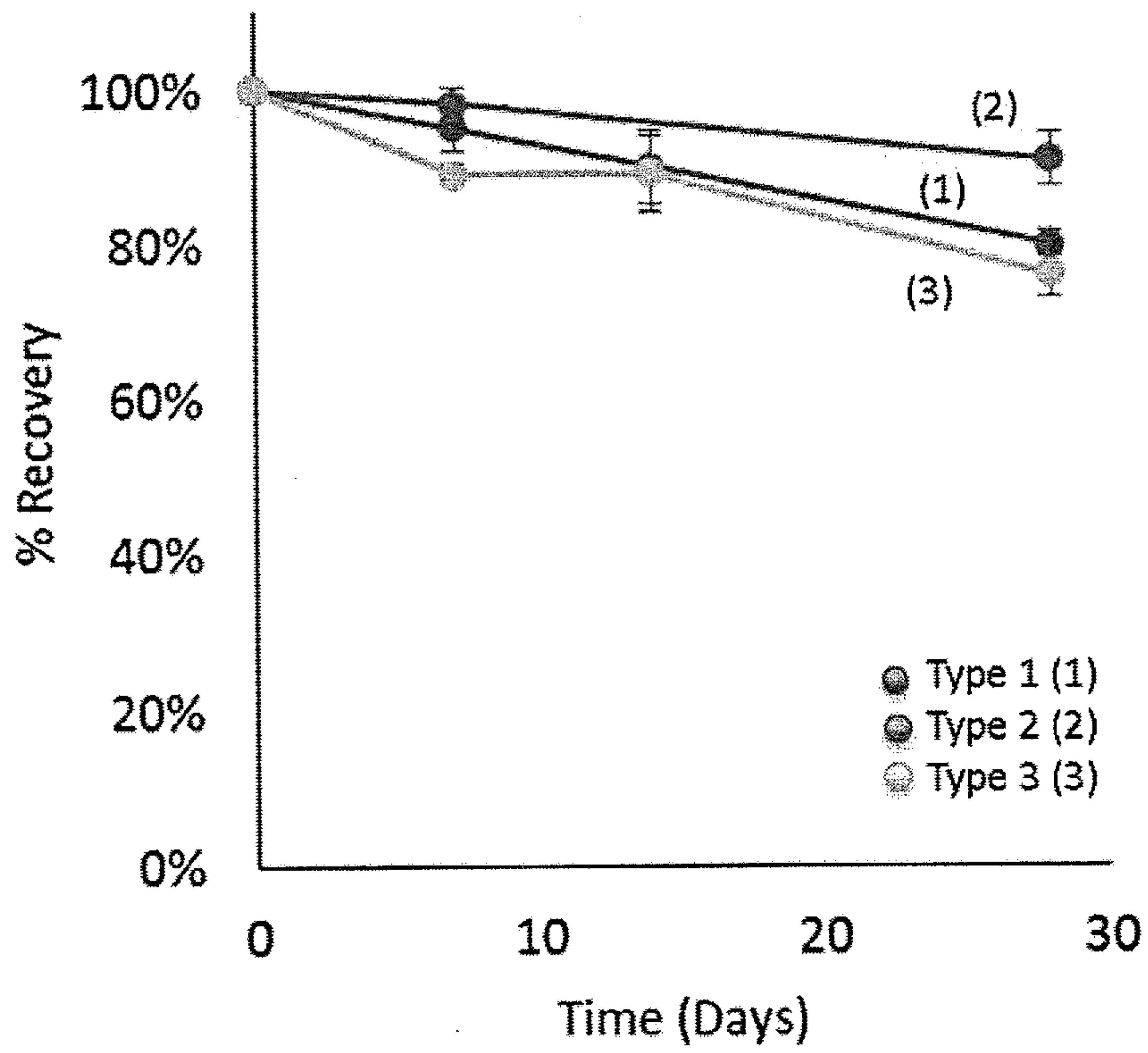
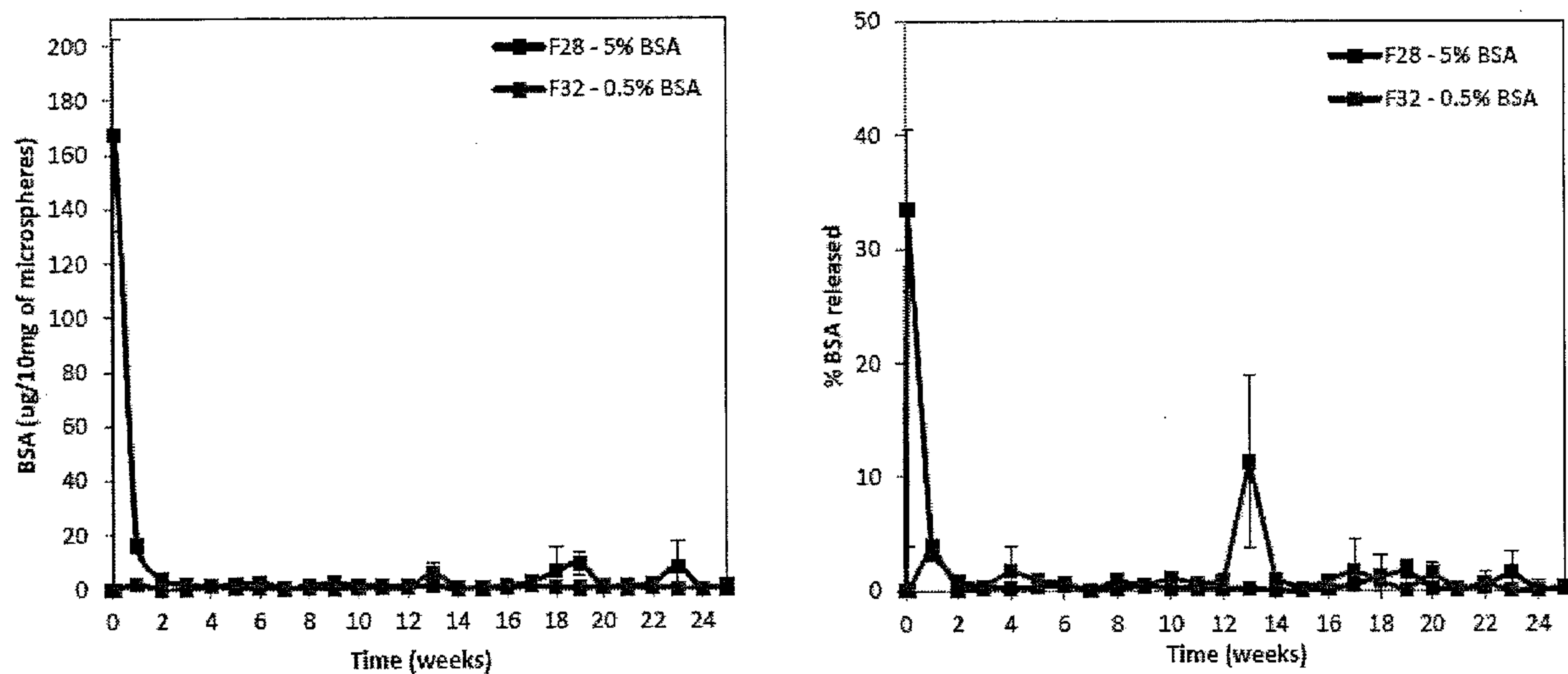


FIG. 2

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## Model protein release kinetics PLLA – 50k



*Burst peaks:*

F28 – 0, 19, and 23 weeks, ...

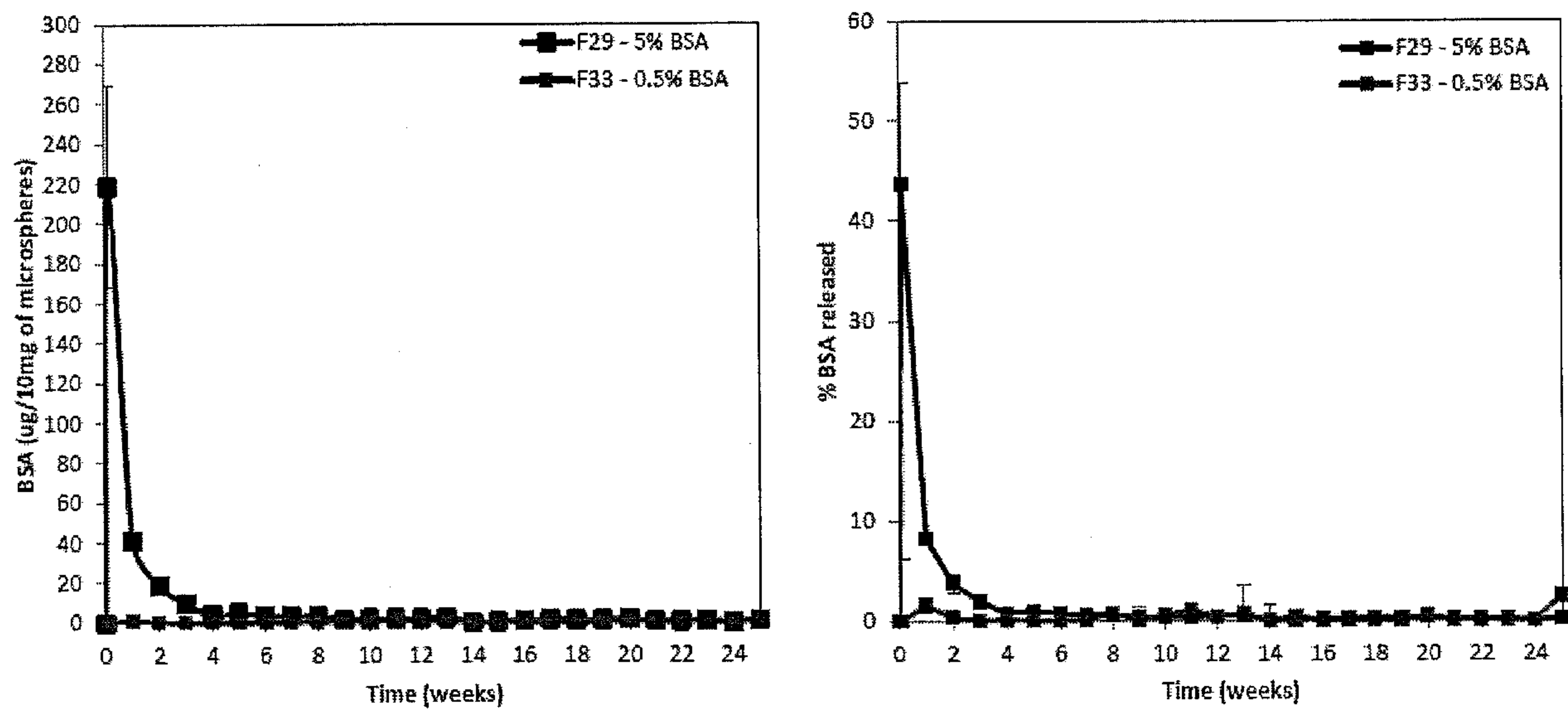
F32 – 1 and 13 weeks, ...

**FIGS. 3A AND 3B**

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# Model protein release kinetics

## PLLA – 100k



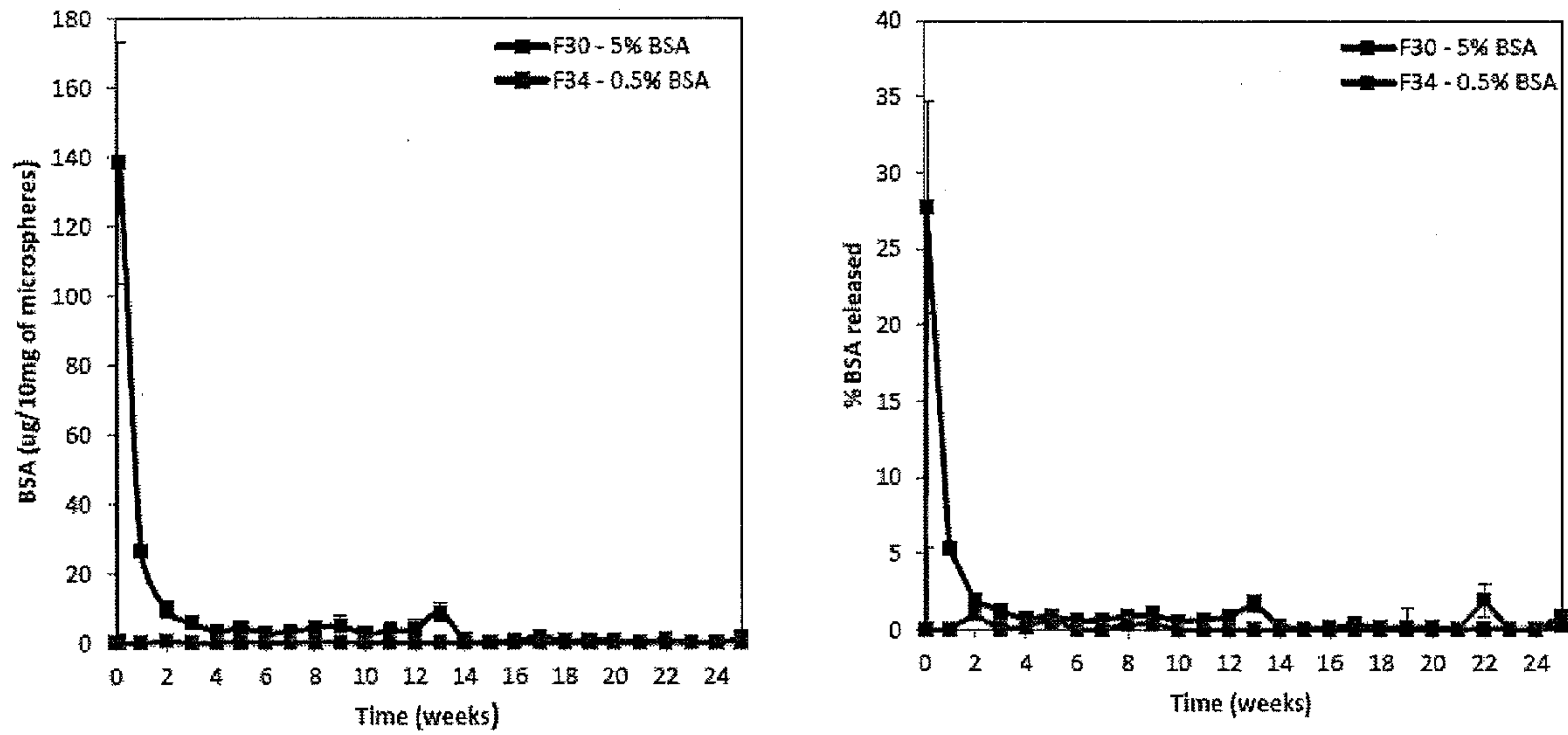
*Burst peaks:*  
F29 – 0 weeks, ...  
F33 – none, ...

FIGS. 4A AND 4B



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## Model protein release kinetics PLLA – 300k

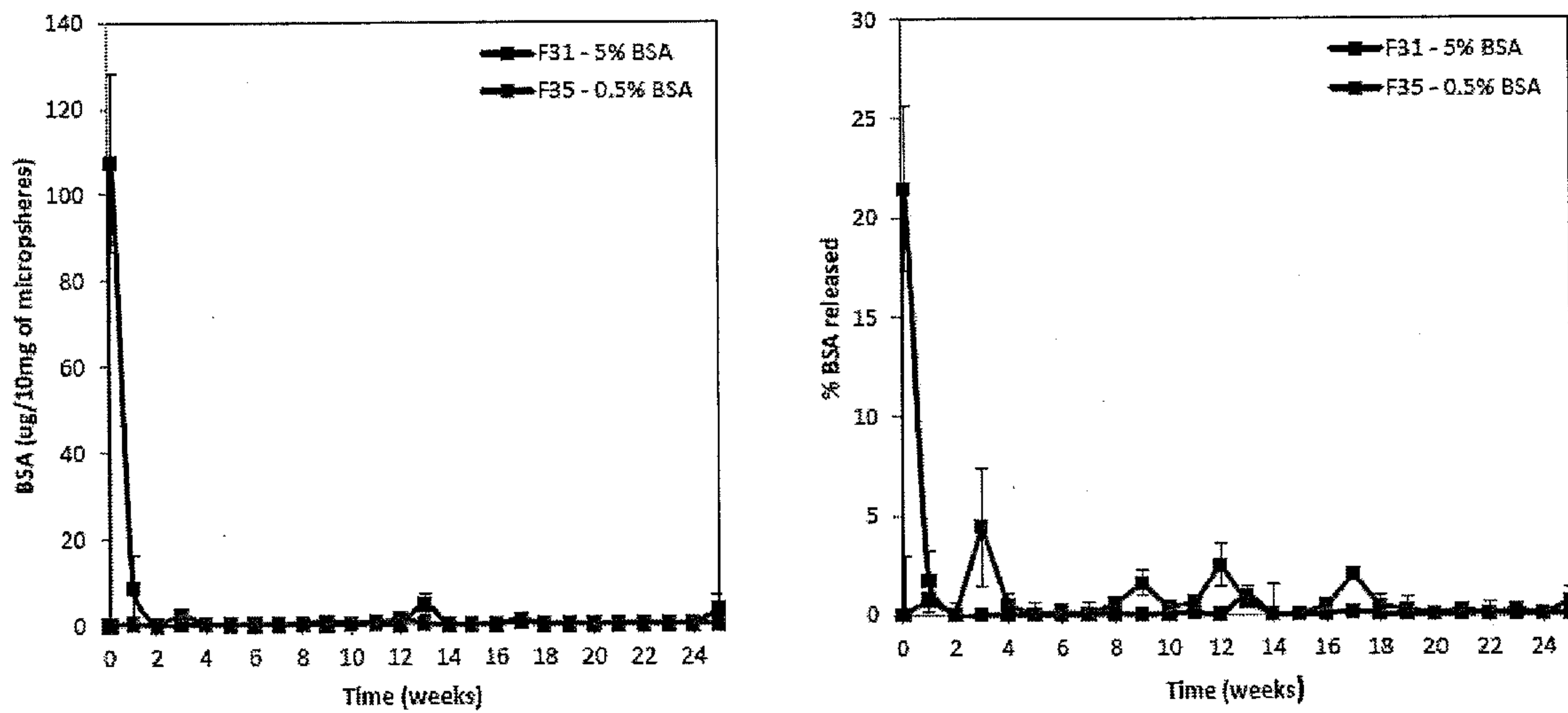


*Burst peaks:*  
F30 – 0 and 13 weeks, ...  
F34 – none, ...

**FIGS 5A AND 5B**

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## Model protein release kinetics P(d,l)LA – 20k



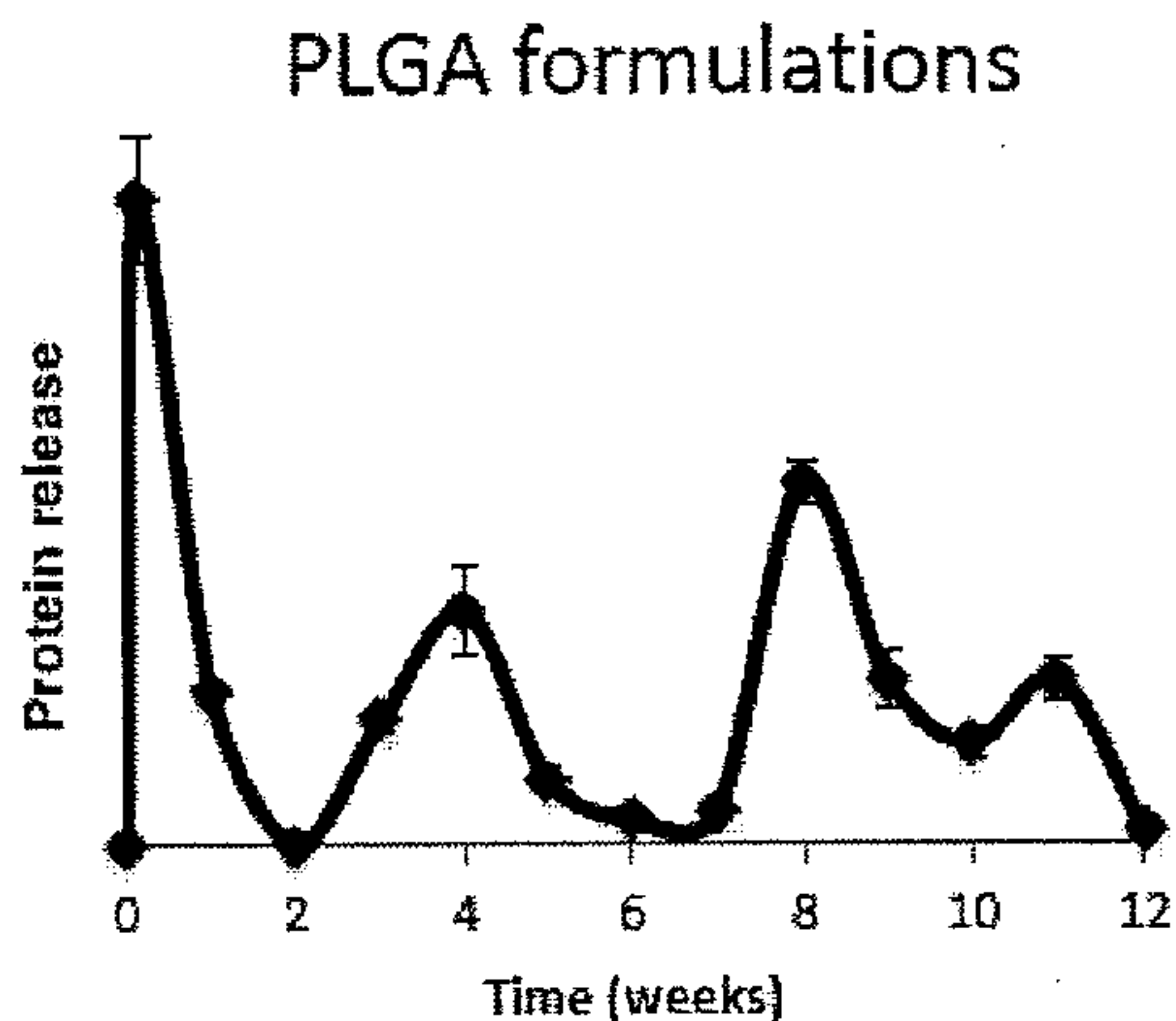
### *Burst peaks:*

F31 – 0 and 13 weeks, ...

F35 – 3, 9, 12, and 17 weeks, ...

**FIGS. 6A AND 6B**

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FIGS. 7A

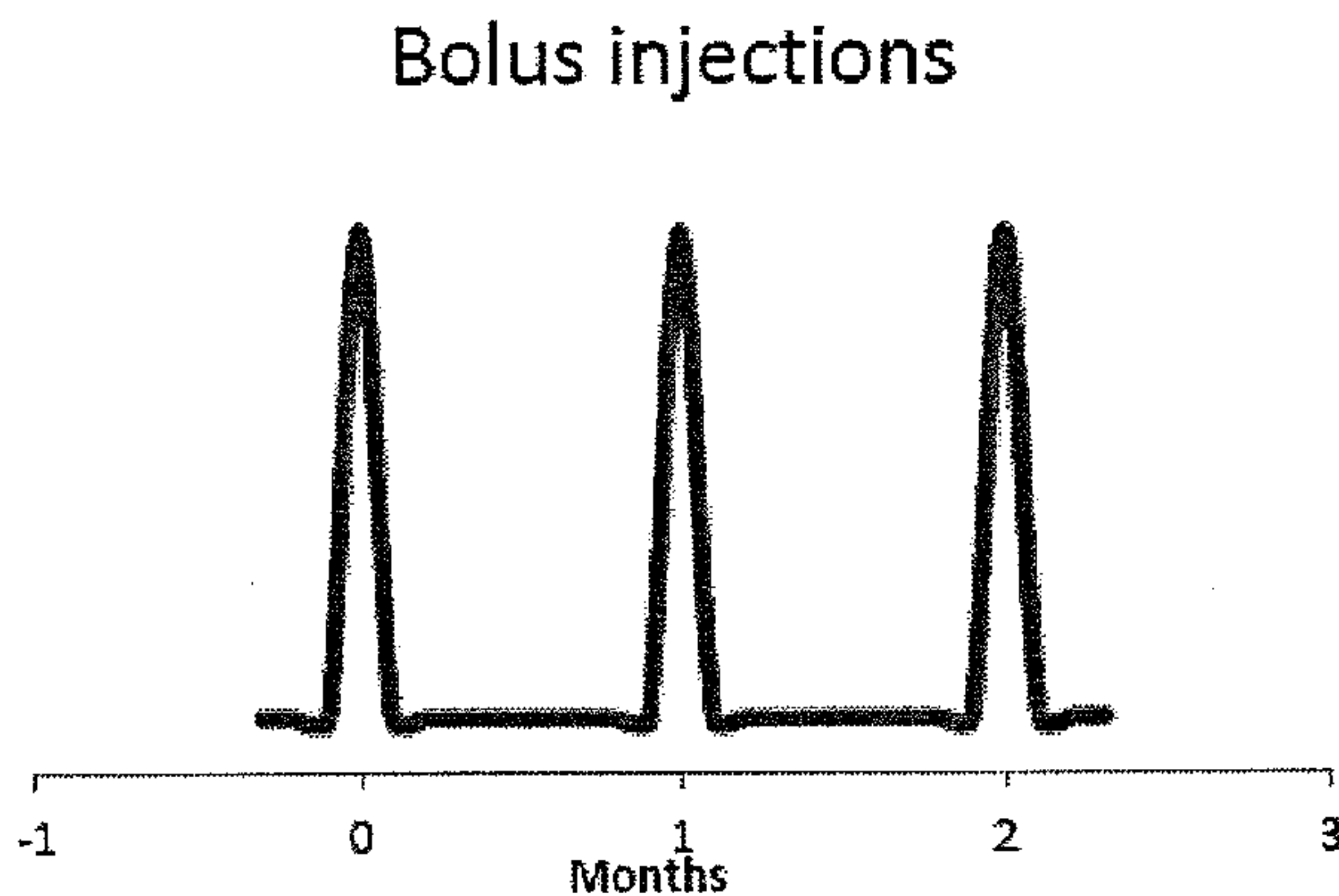
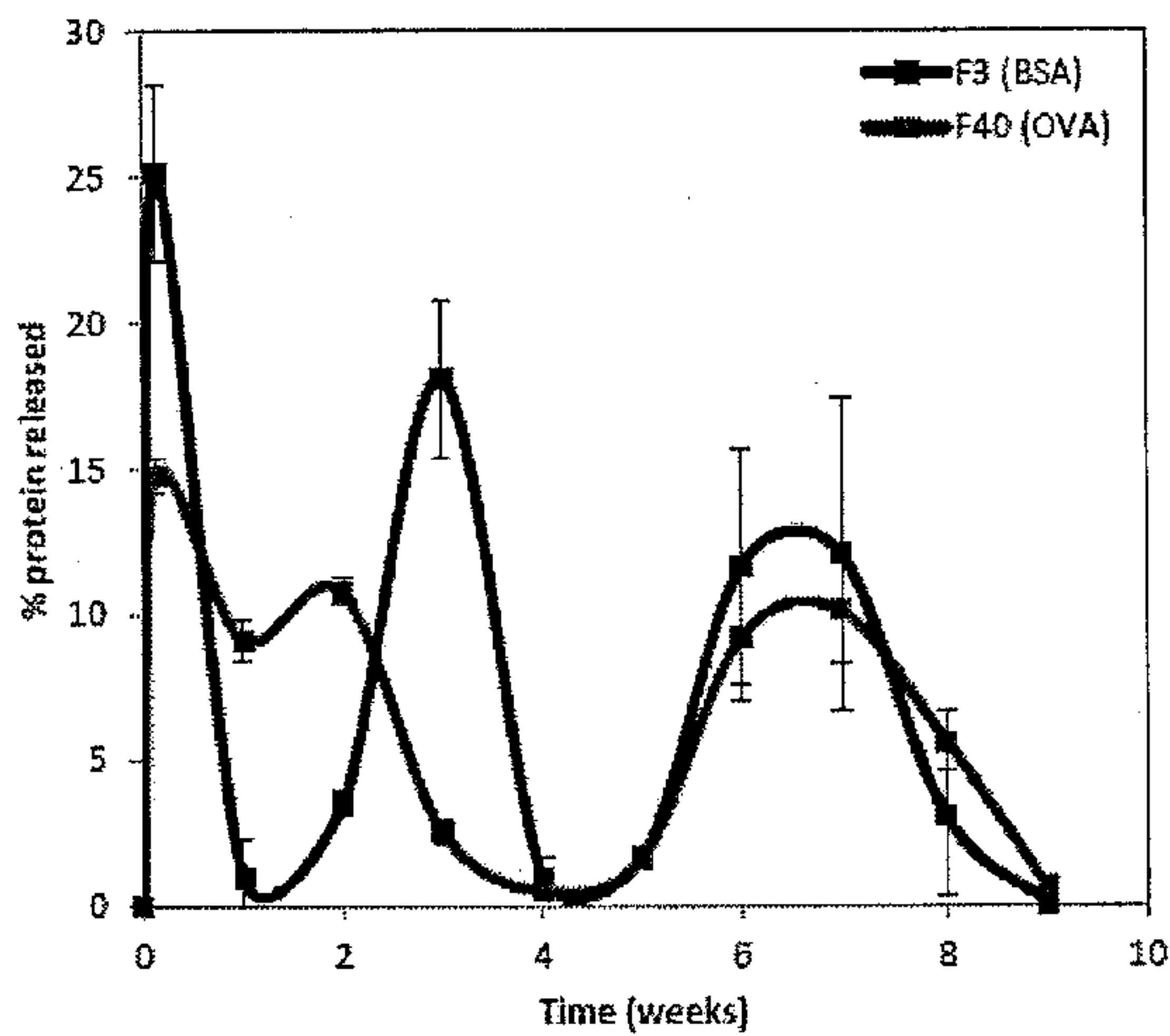
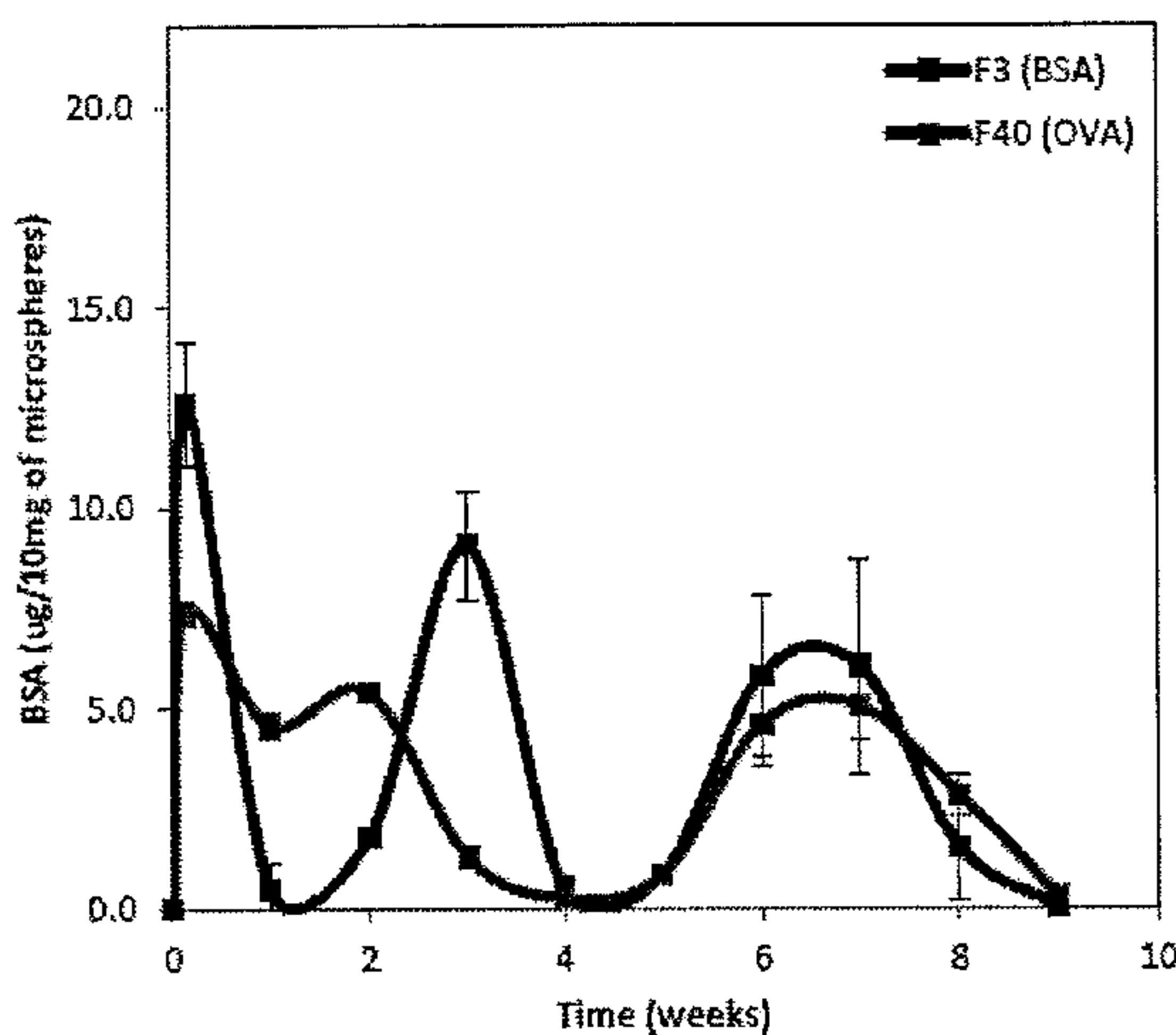


FIG. 7B

Model protein release kinetics  
 PLGA-COOH - 50/50 – 20k, 0.5% Loading

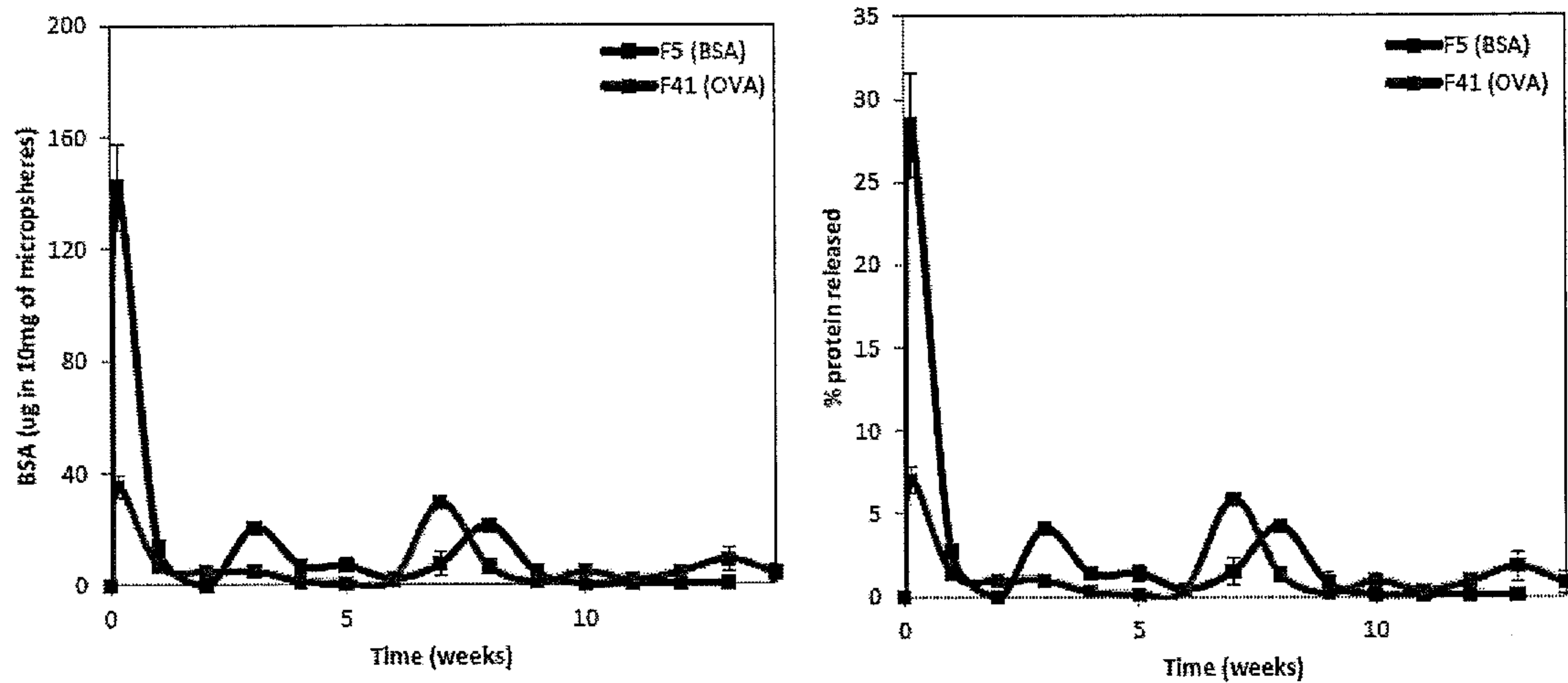


*Burst peaks:*  
 F3 – 0, 3, and 6.5 weeks  
 F40 – 0, 2, and 6.5 weeks

FIGS. 8A AND 8B

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## Model protein release kinetics PLGA-COOH - 50/50 – 31k, 5% Loading



*Burst peaks:*

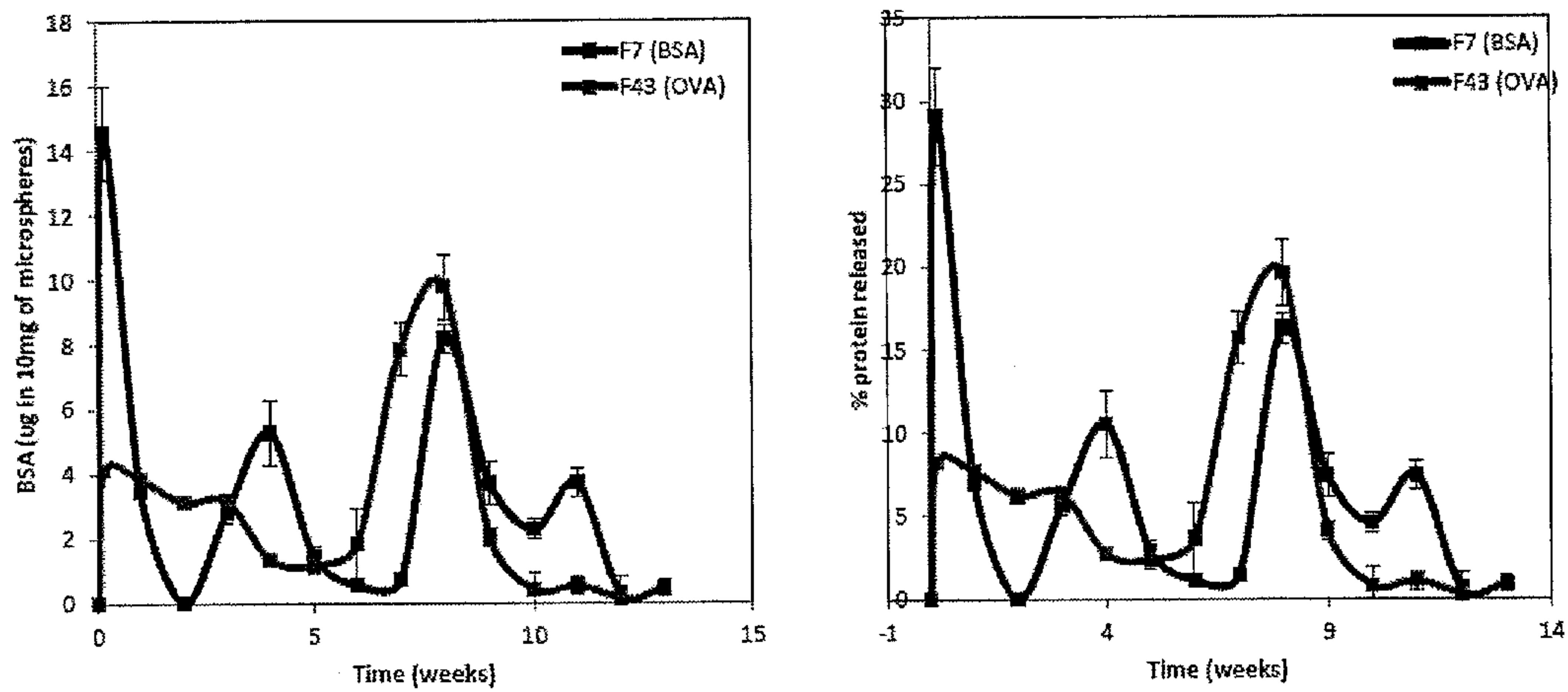
F5 – 0, 3, and 8 weeks

F41 – 0, 3, 7, and 13 weeks

**FIGS. 9A AND 9B**

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Model protein release kinetics  
PLGA-COOH - 50/50 – **31k, 0.5% Loading**



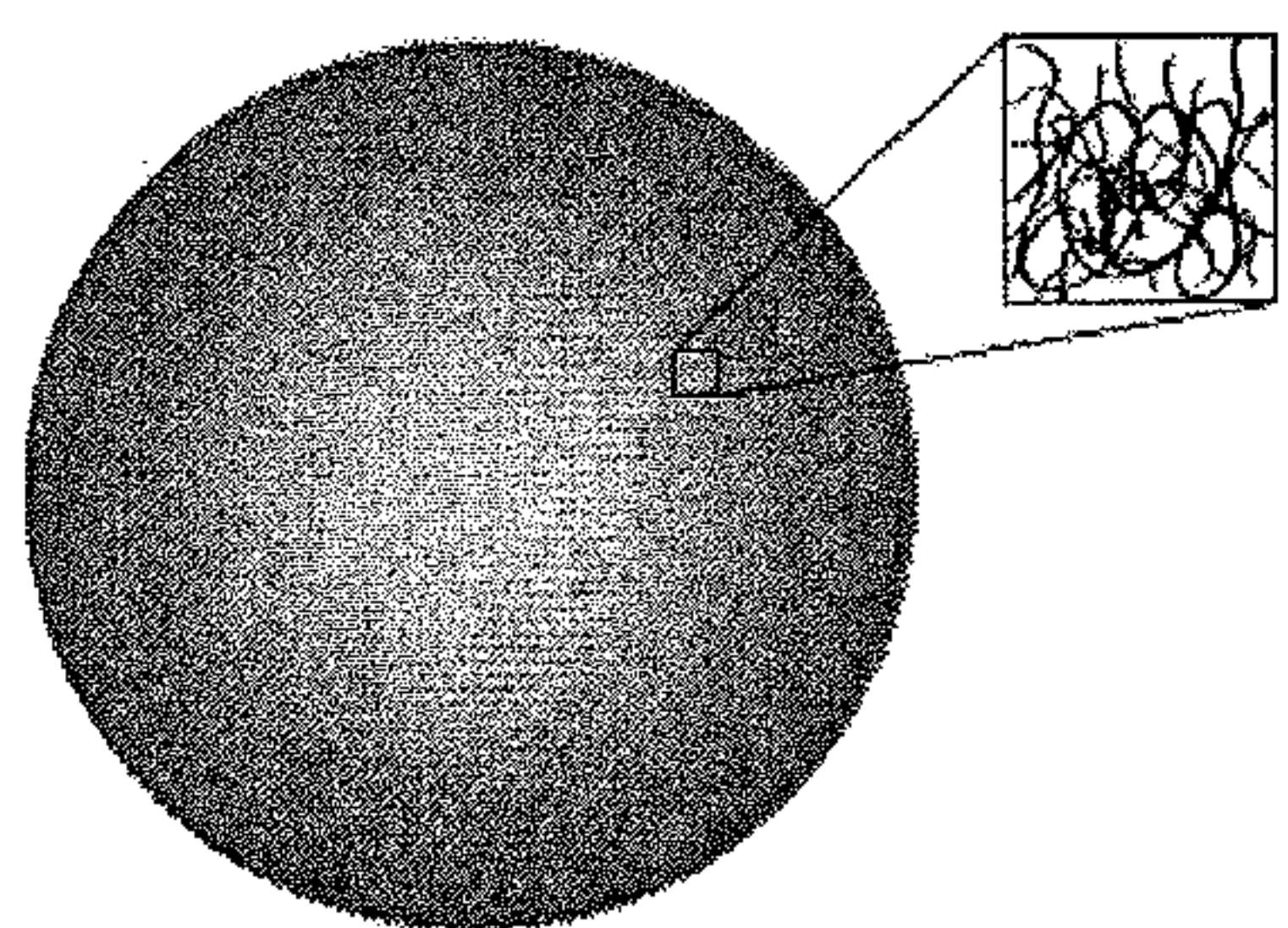
*Burst peaks:*

F7 – 0, 4, 8, and 11 weeks

F43 – 0, 3, 8 and 11 weeks

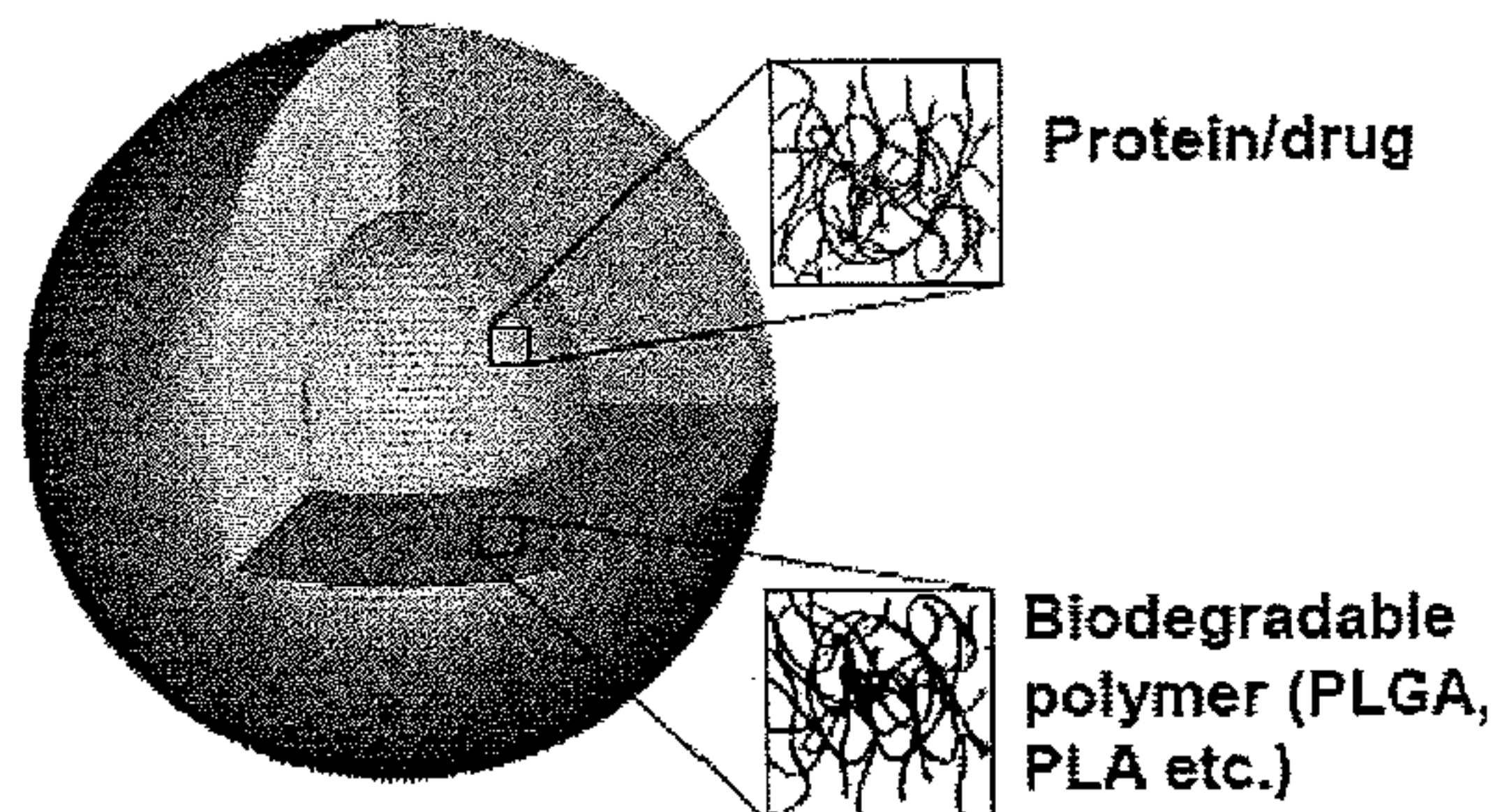
**FIGS. 10A AND 10B**

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Emulsion method

FIG. 11A



3D printing

FIG. 11B

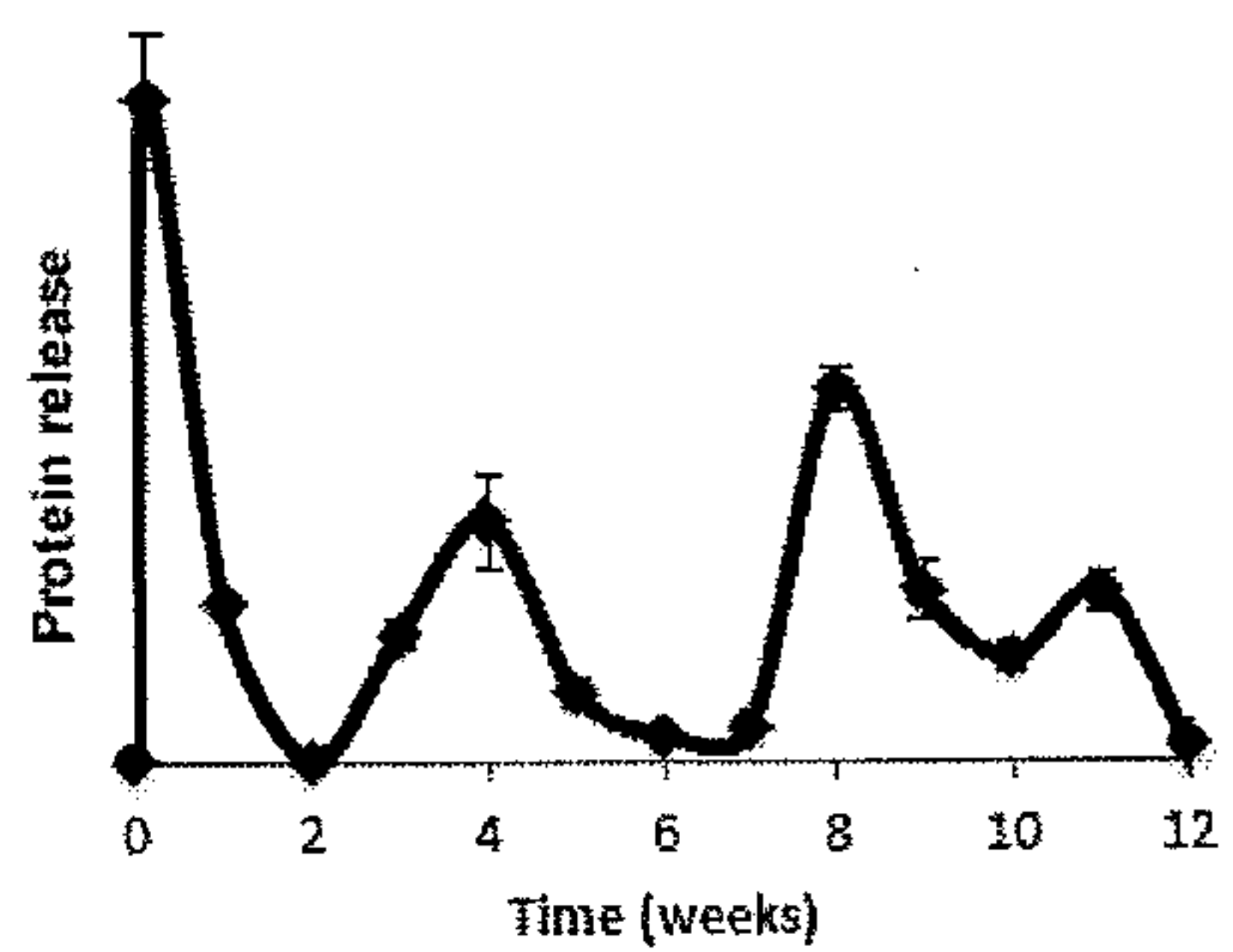


FIG. 11C

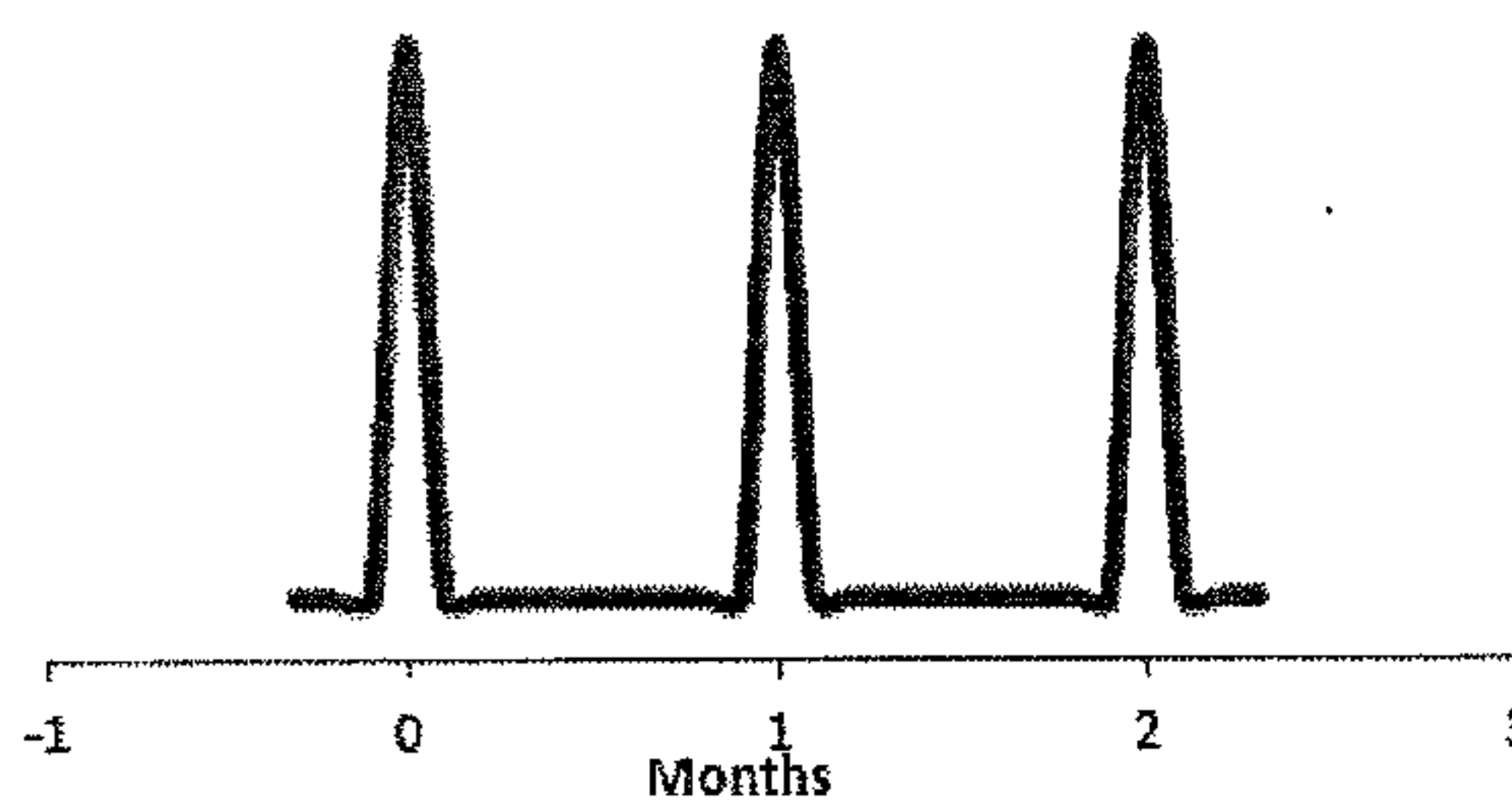


FIG. 11D

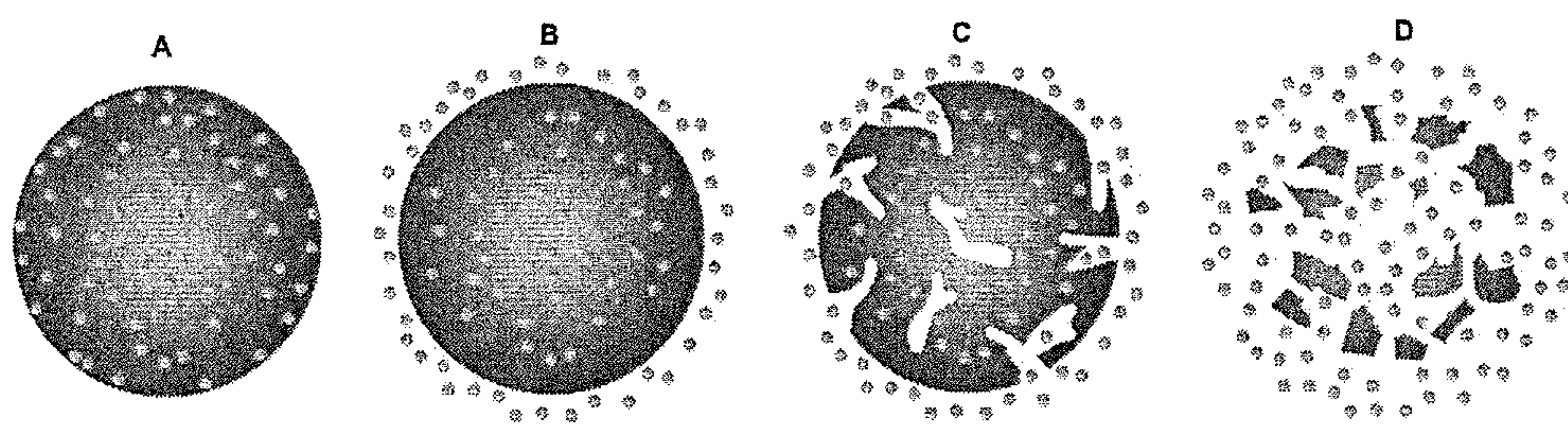
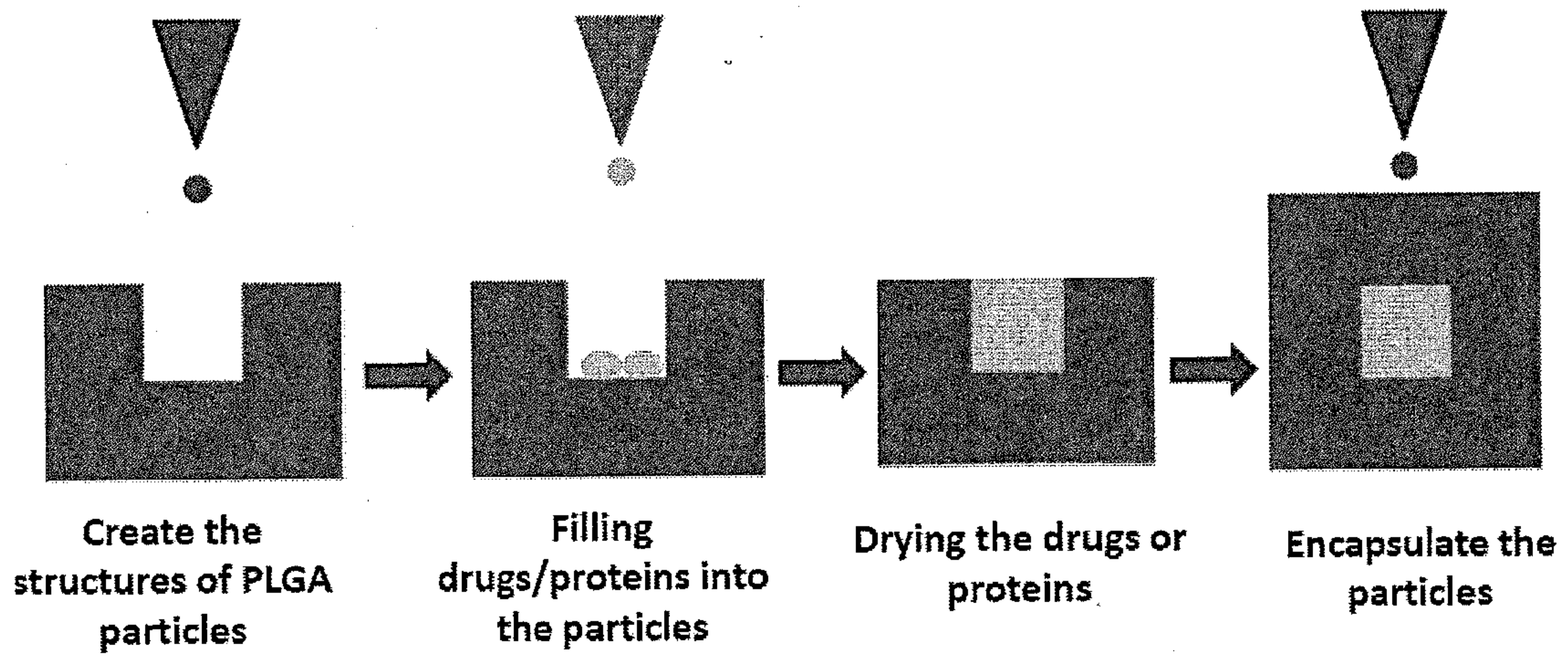


FIG. 11E

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## 3D-printing Encapsulation Process



*First optimize printing in the x, y, and z directions by printing cube-shaped PLGA structures filled with vaccine.*

FIG. 12A

FIG. 12B

FIG. 12C

FIG. 12D

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# Piezoelectric nozzle – jetting waveform

*The waveform parameters have to be optimized to produce uniform, single droplets of ink during printing.*

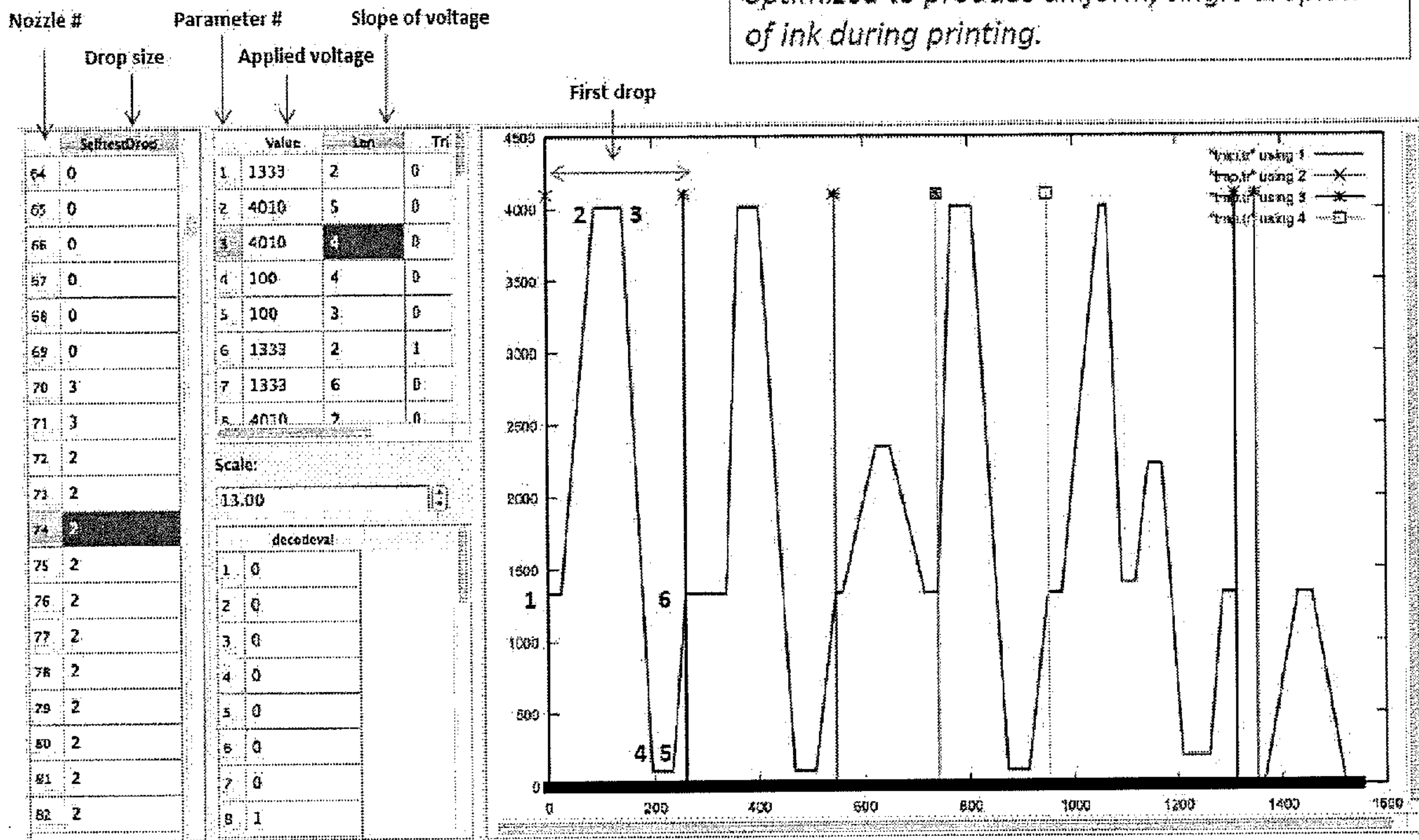


FIG. 13



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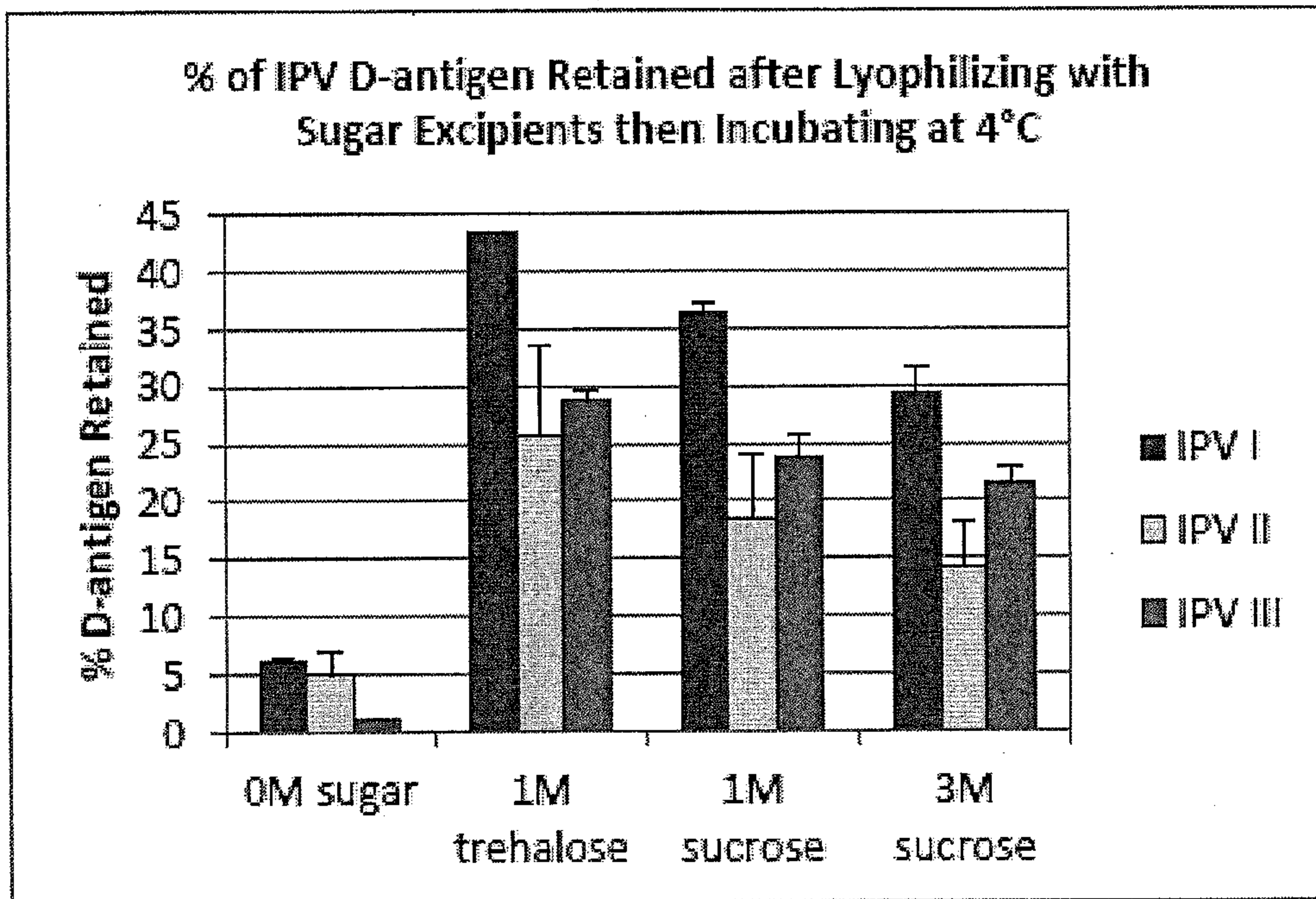


FIG. 14A

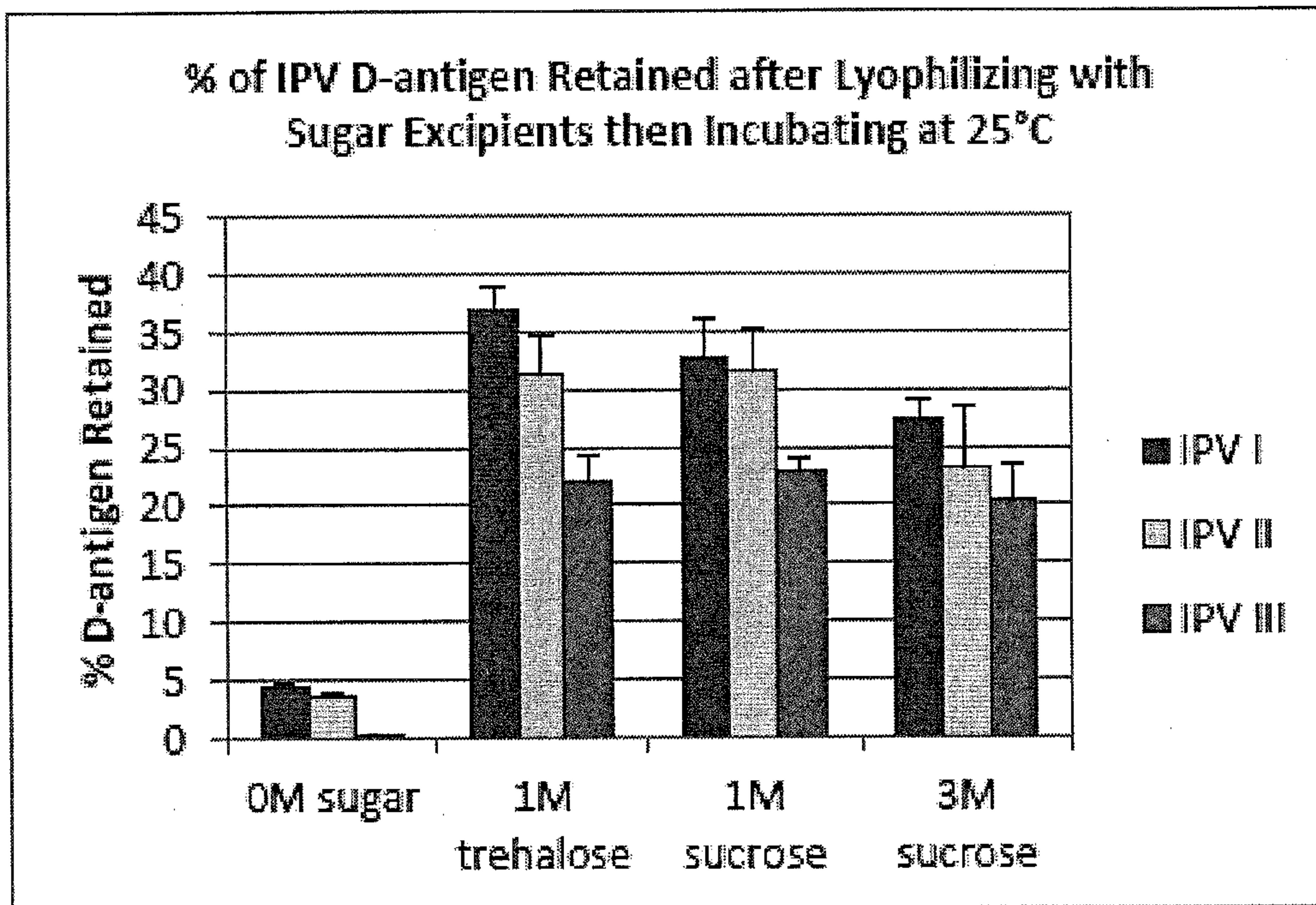


FIG. 14B

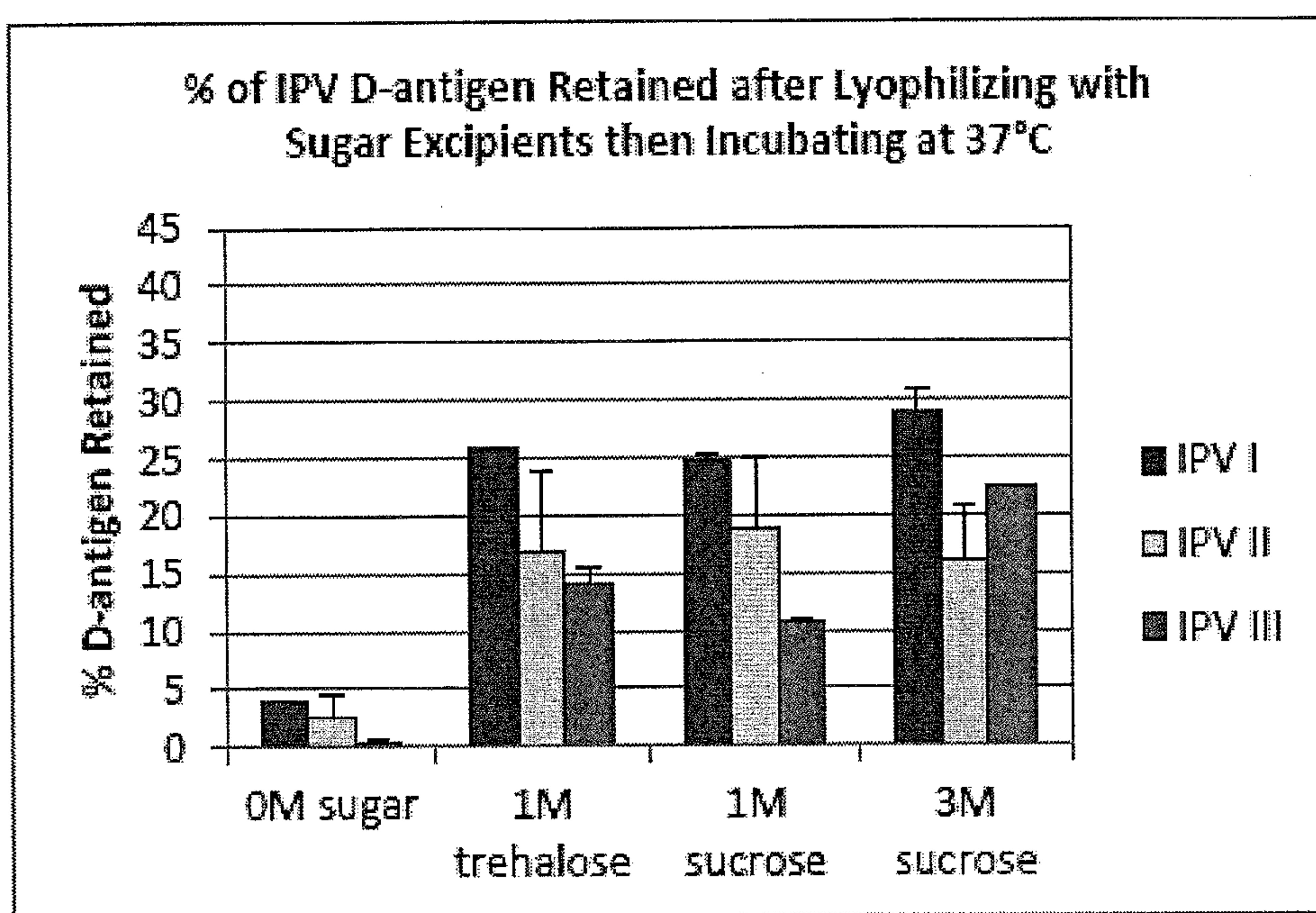


FIG. 14C

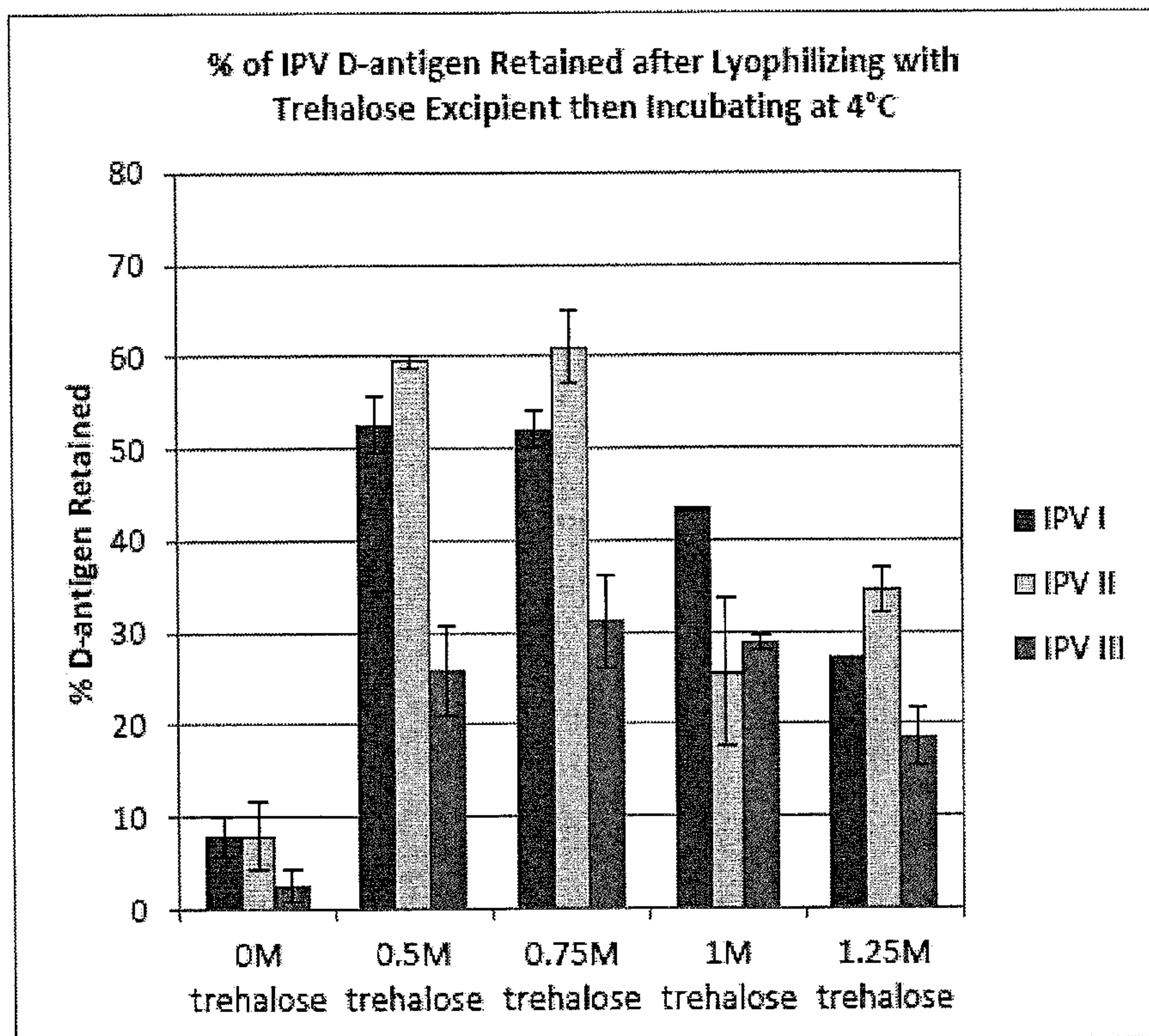


FIG. 15A

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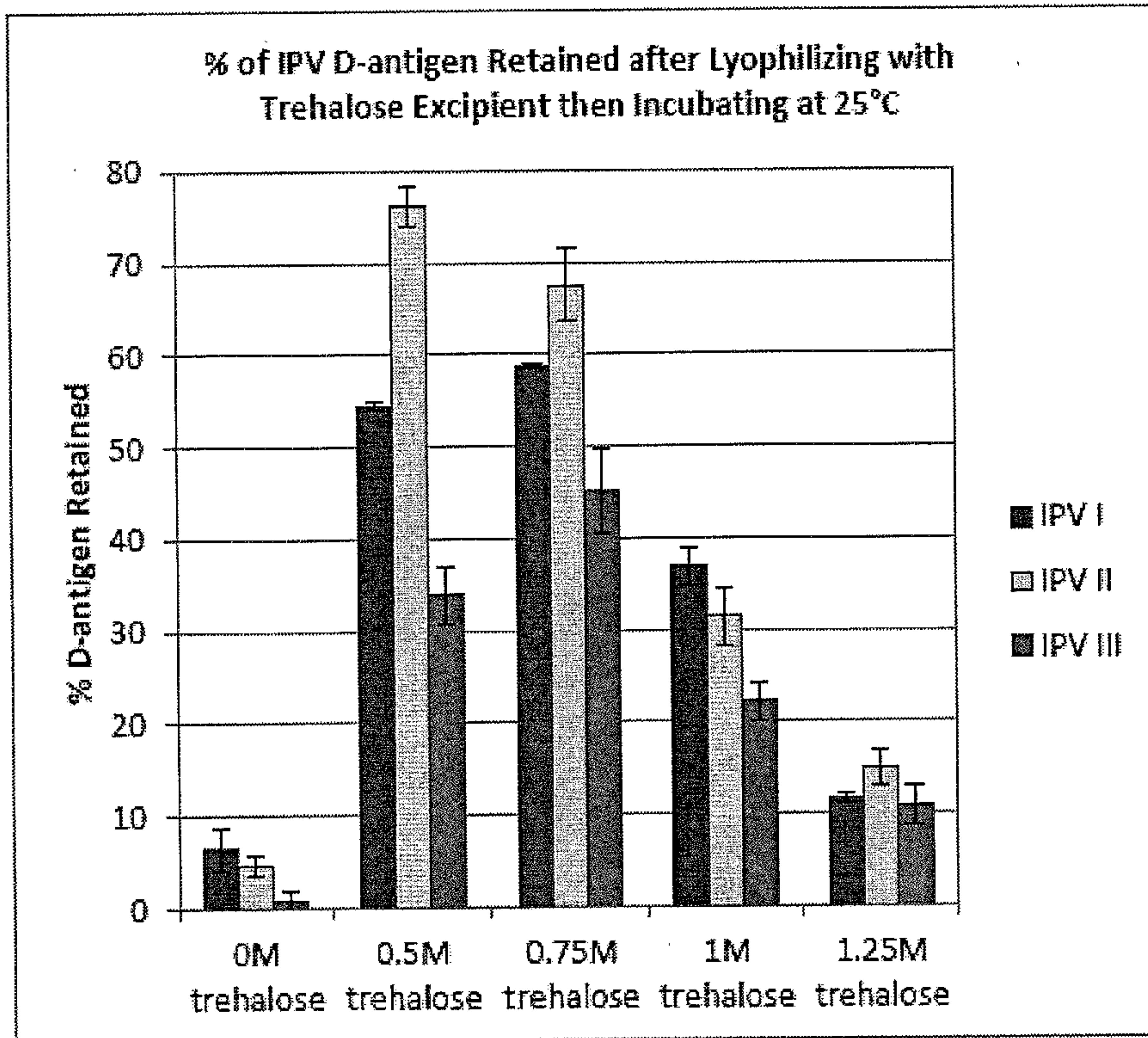


FIG. 15B

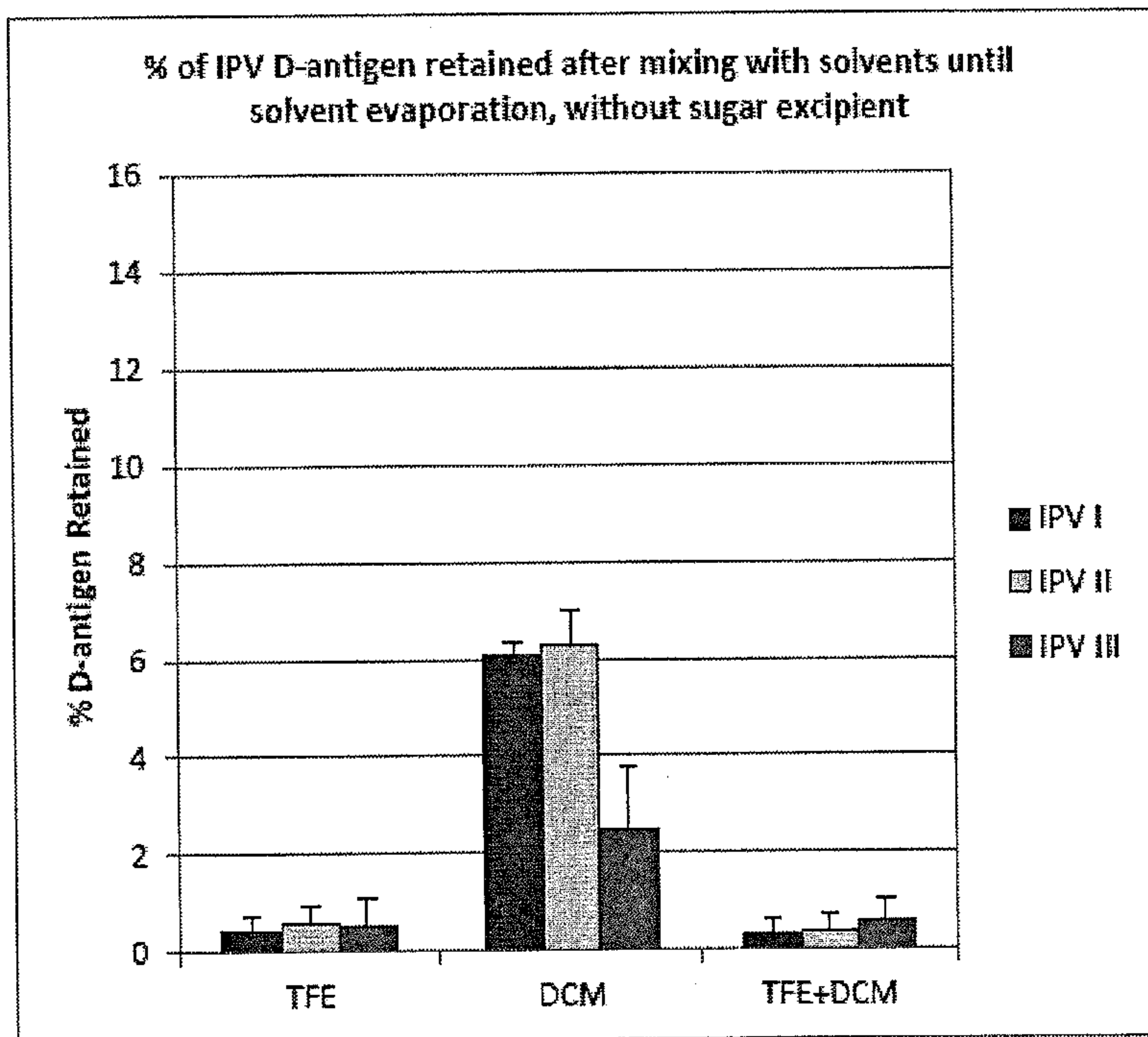


FIG. 16A

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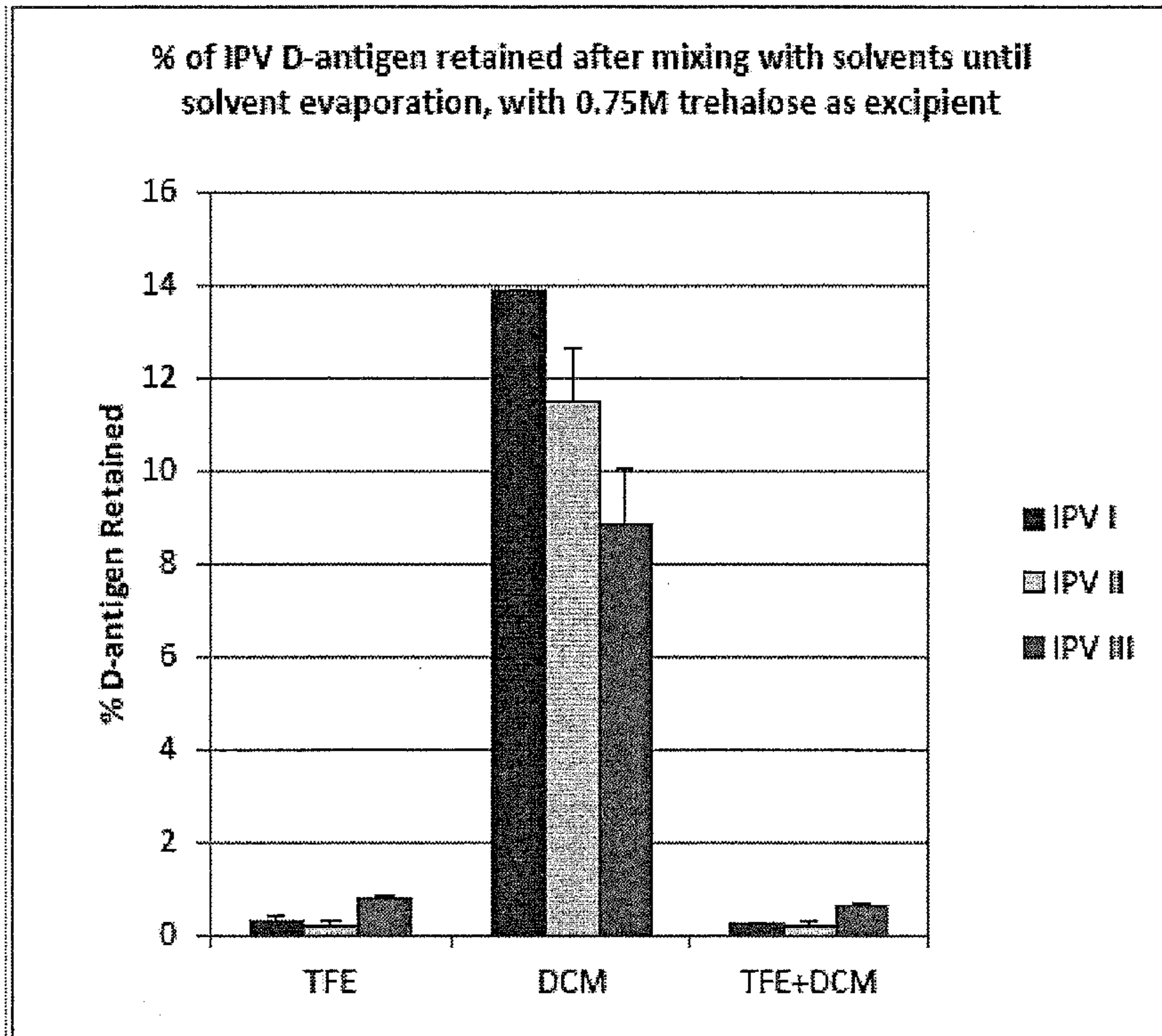


FIG. 16B

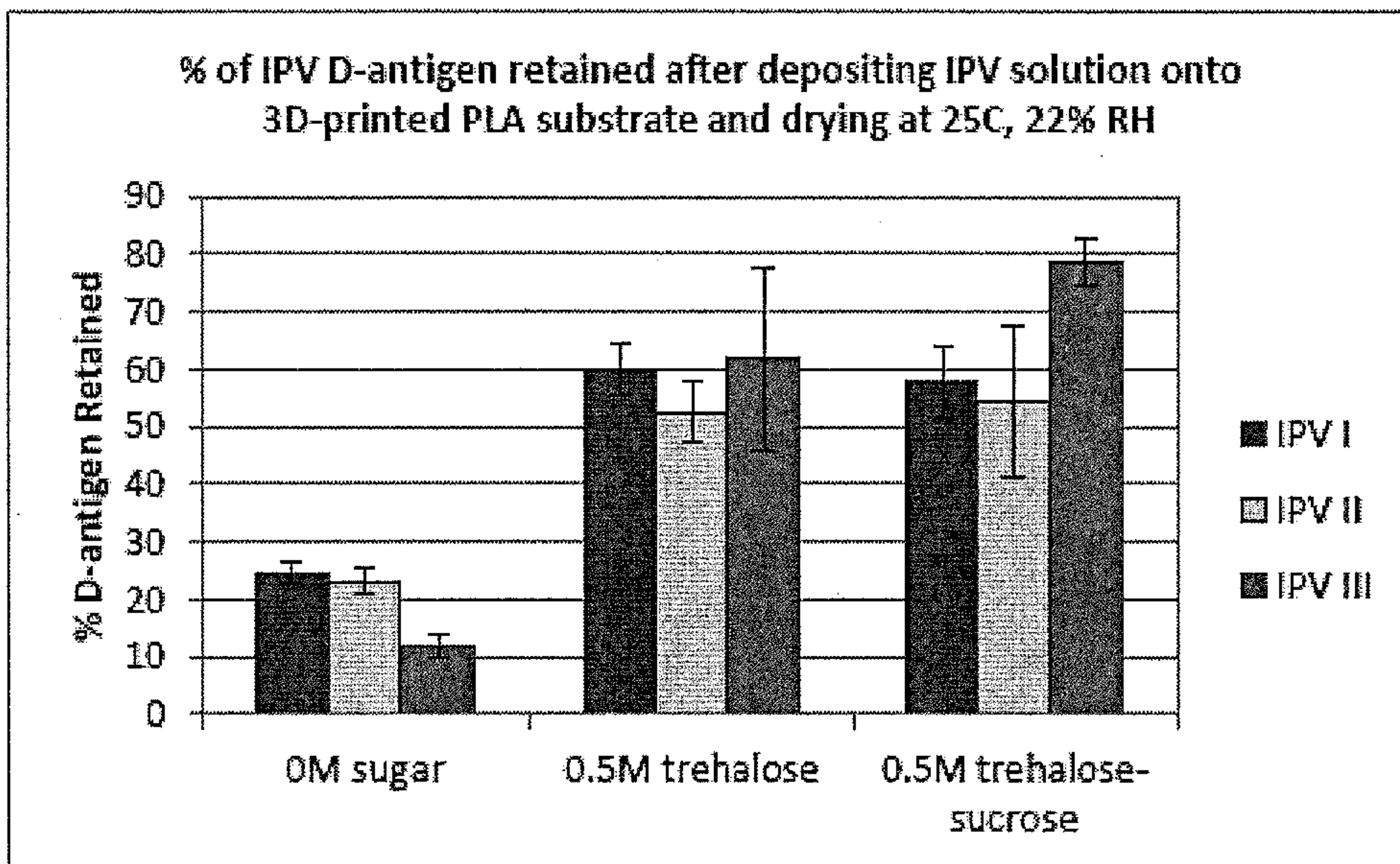


FIG. 17A

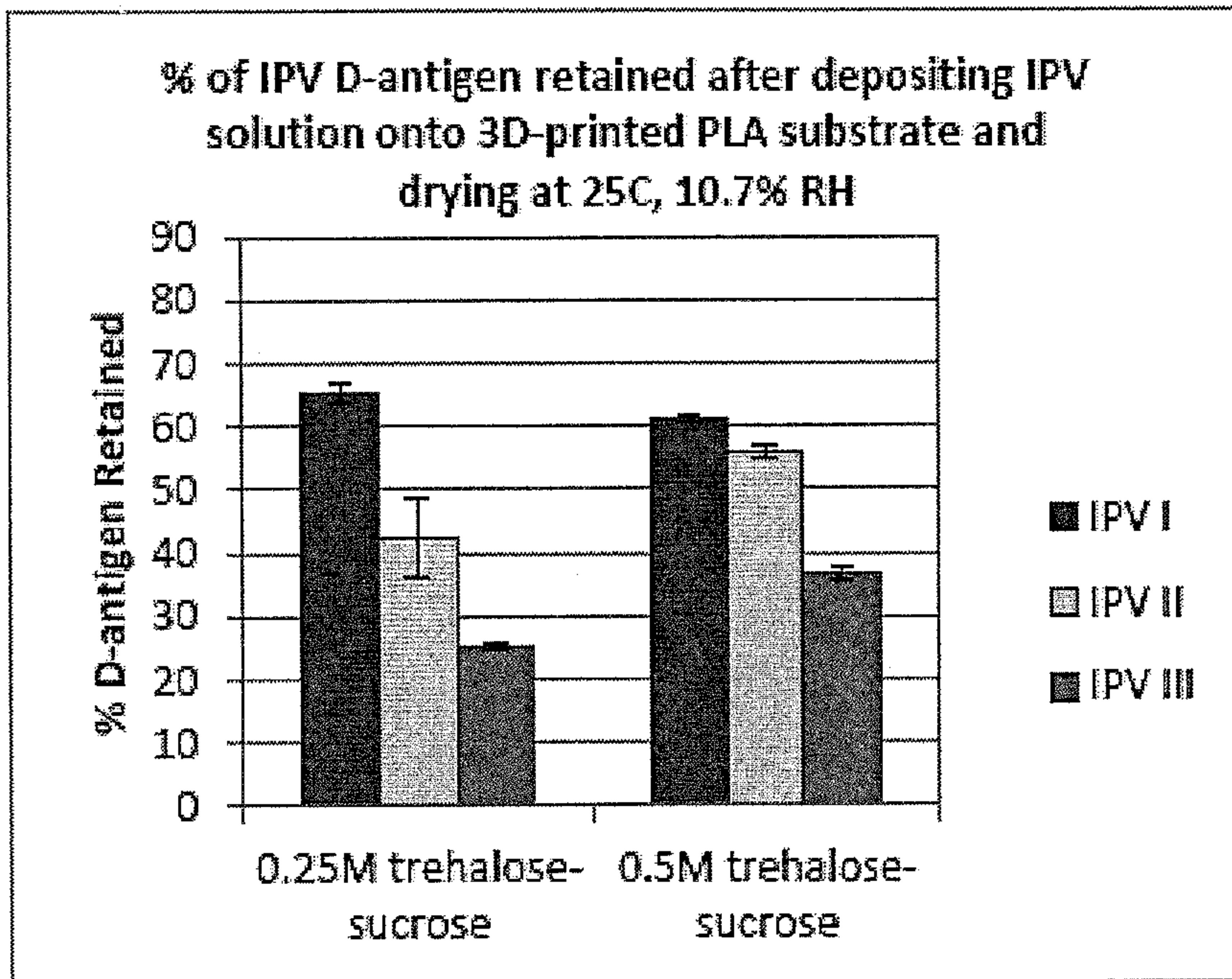


FIG. 17B

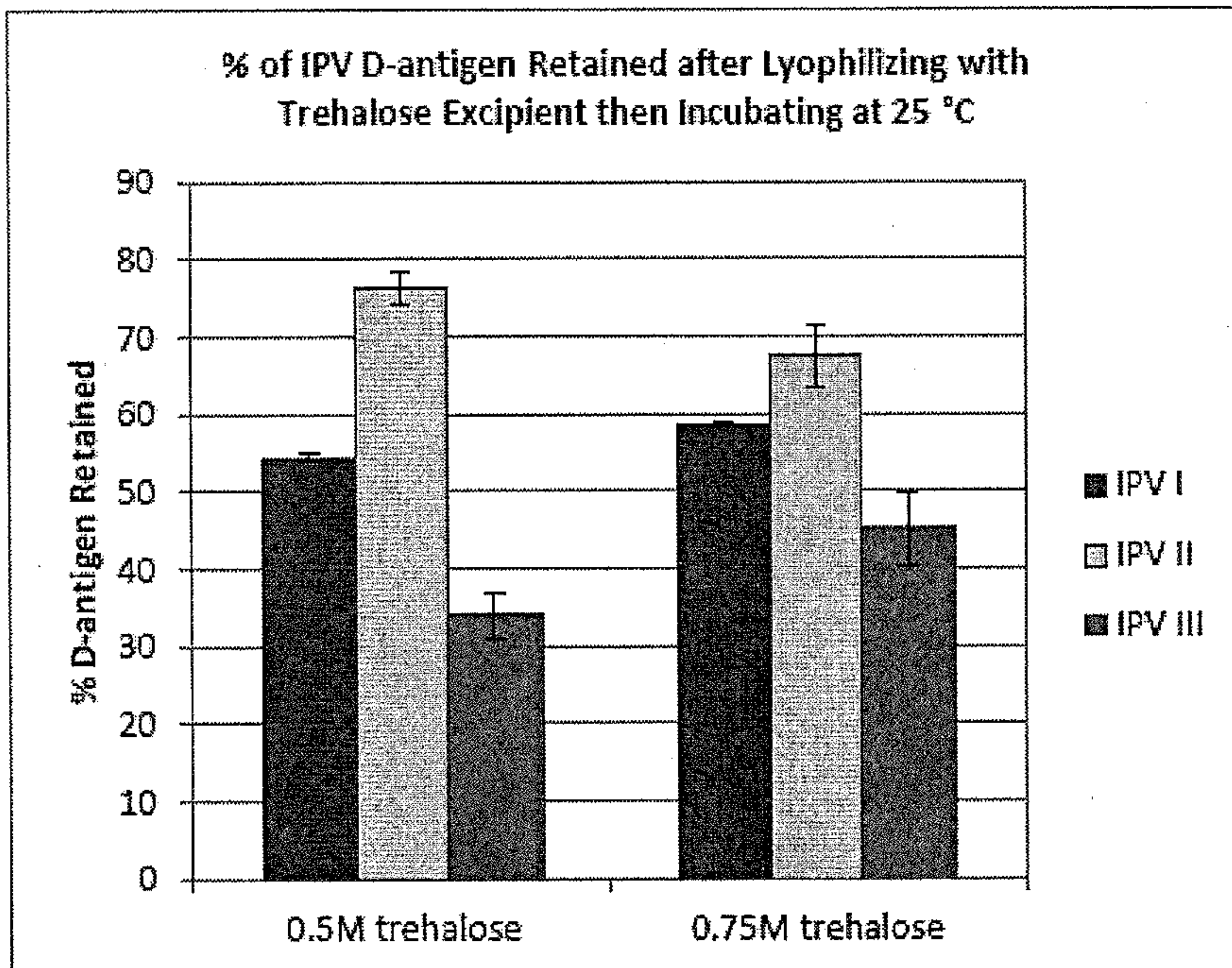
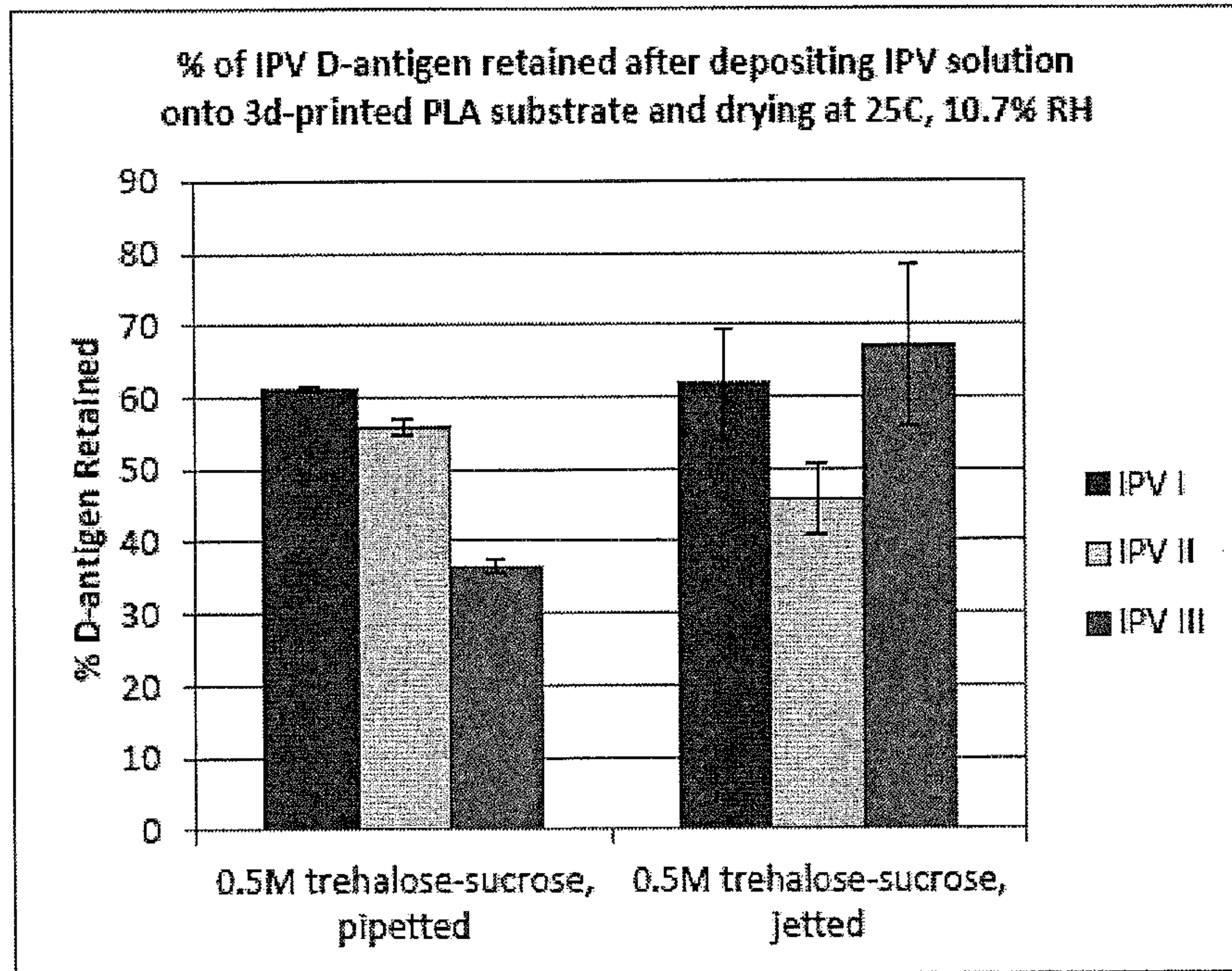


FIG. 18A

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**FIG. 18B**

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week 1 serum data

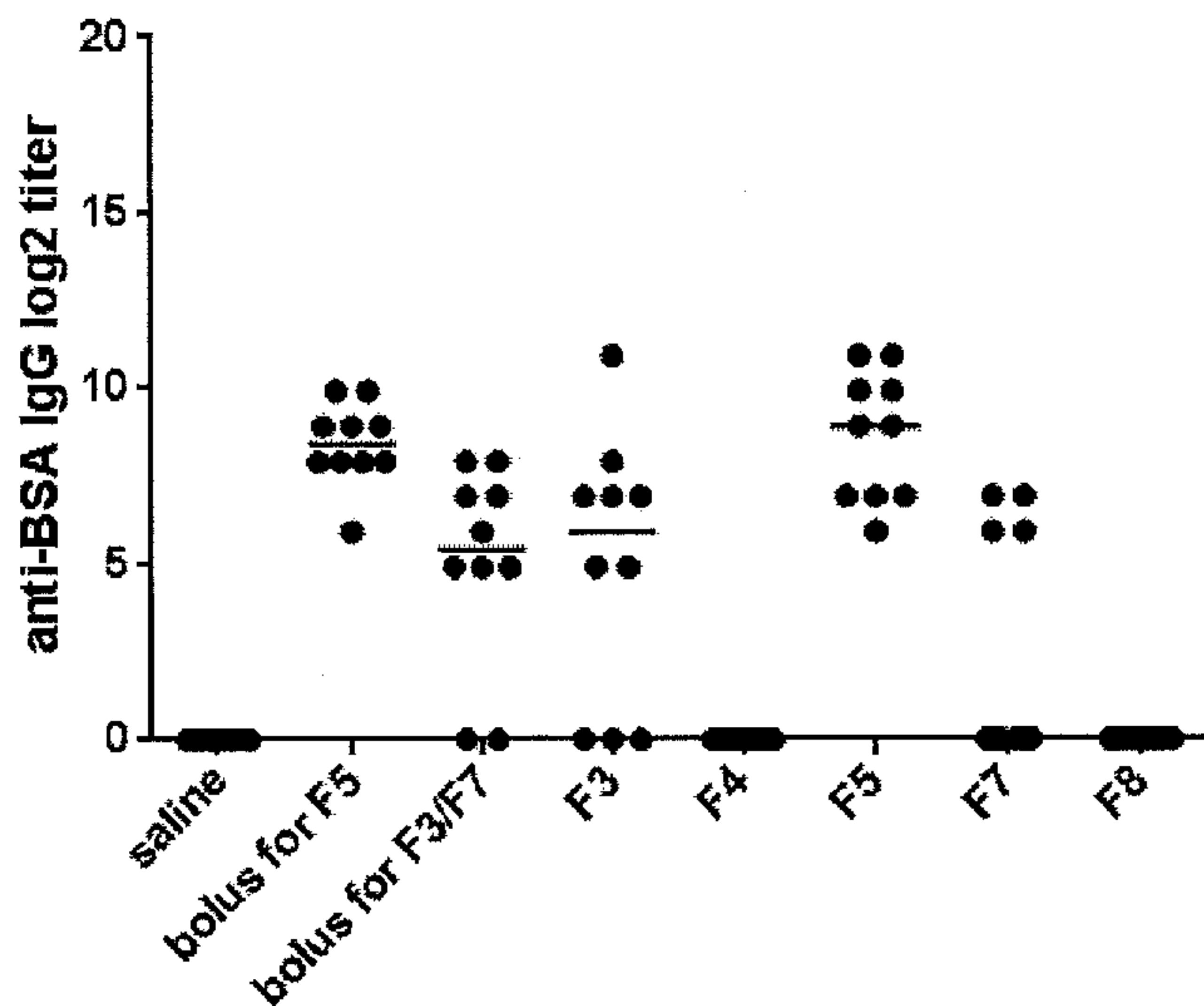


FIG. 19A

week 2 serum data

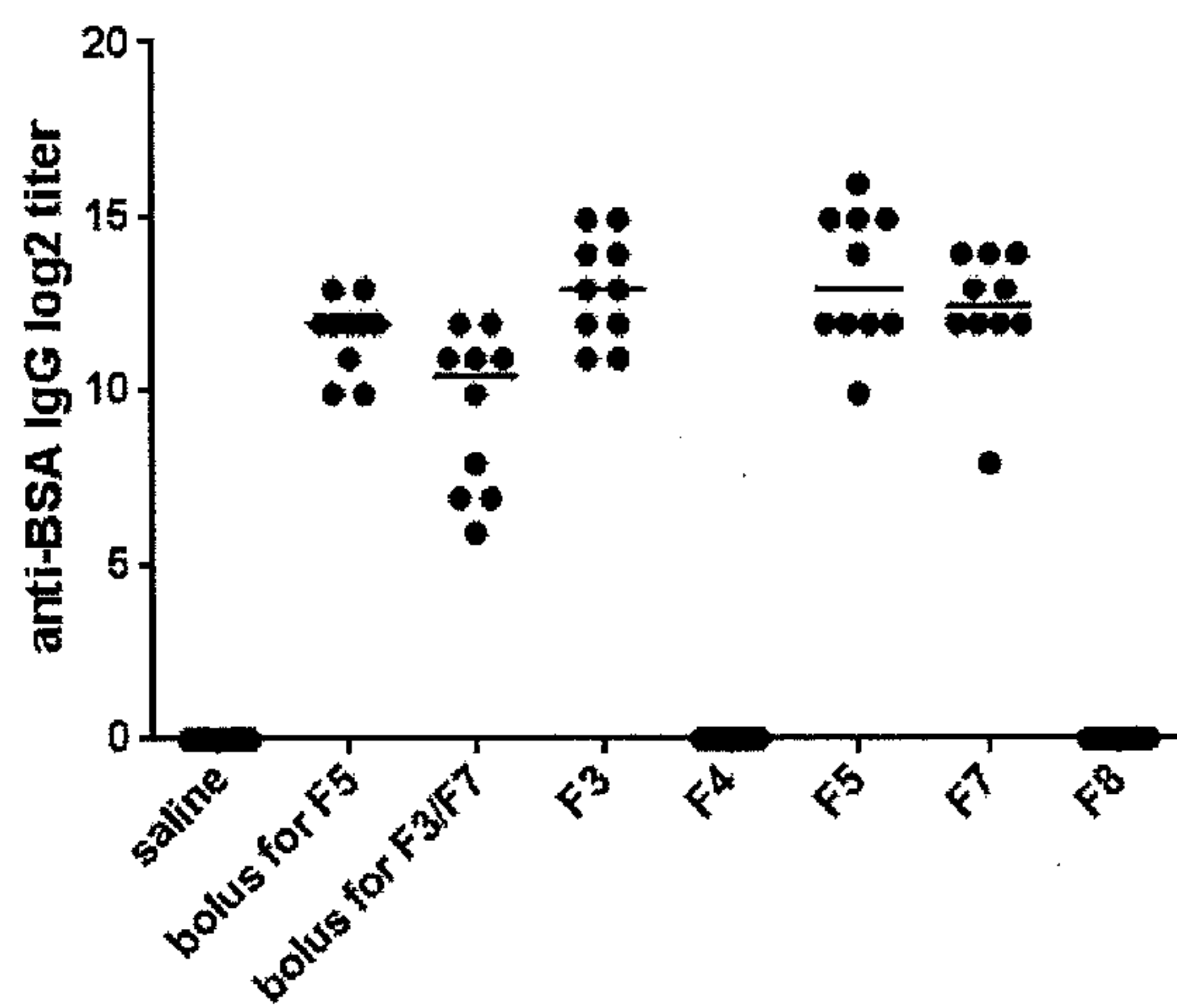


FIG. 19B

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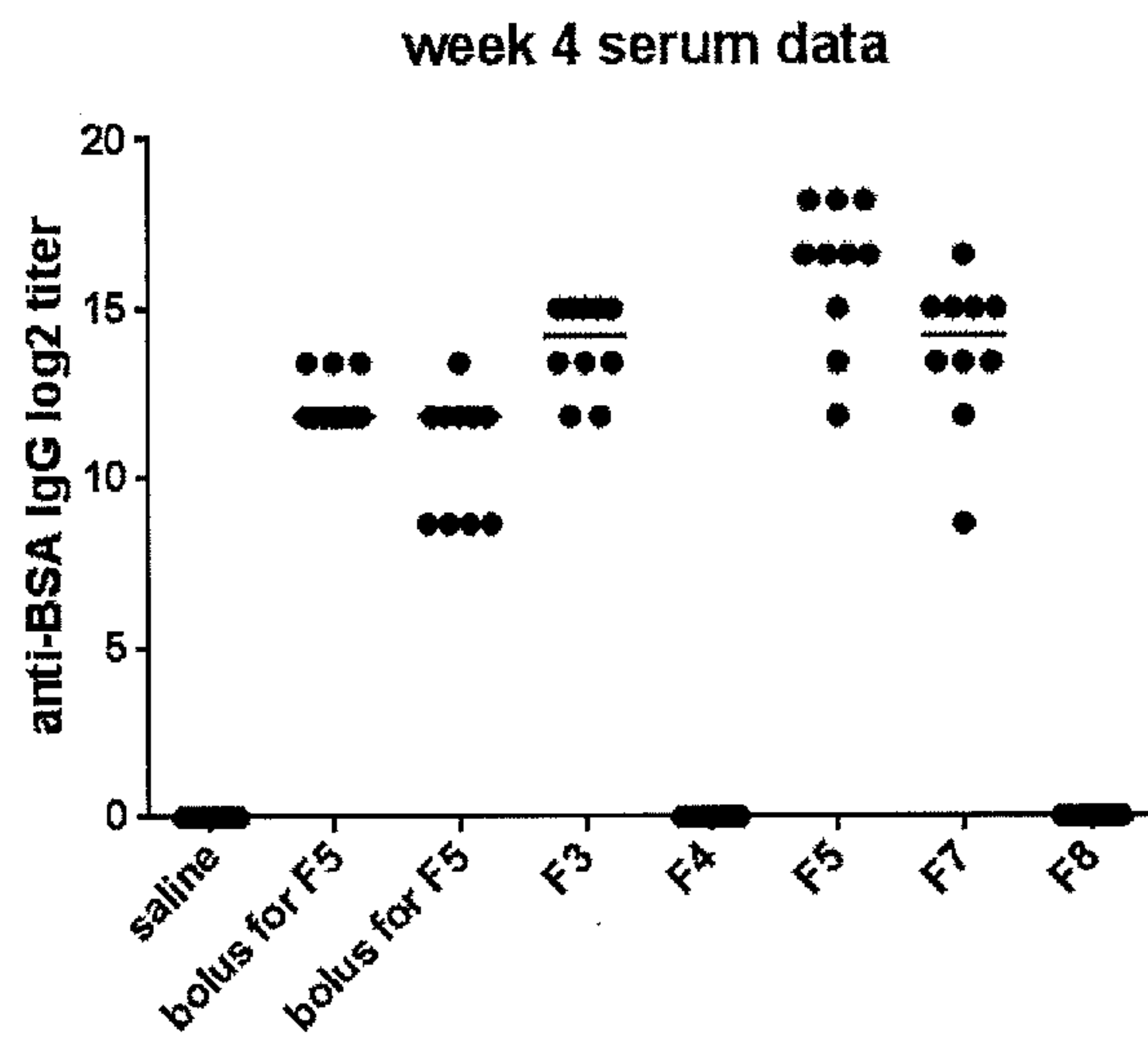


FIG. 19C

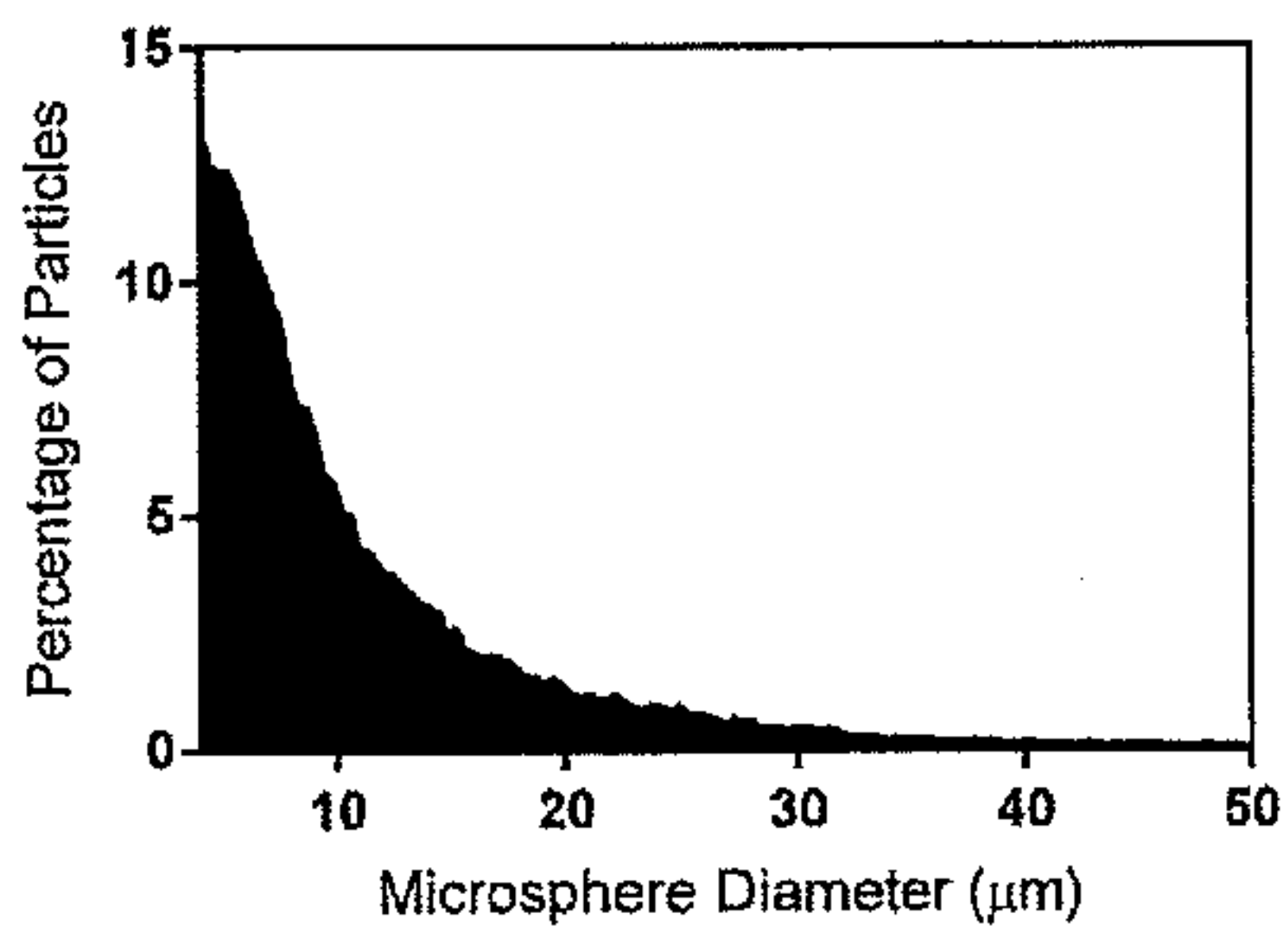


FIG. 20A

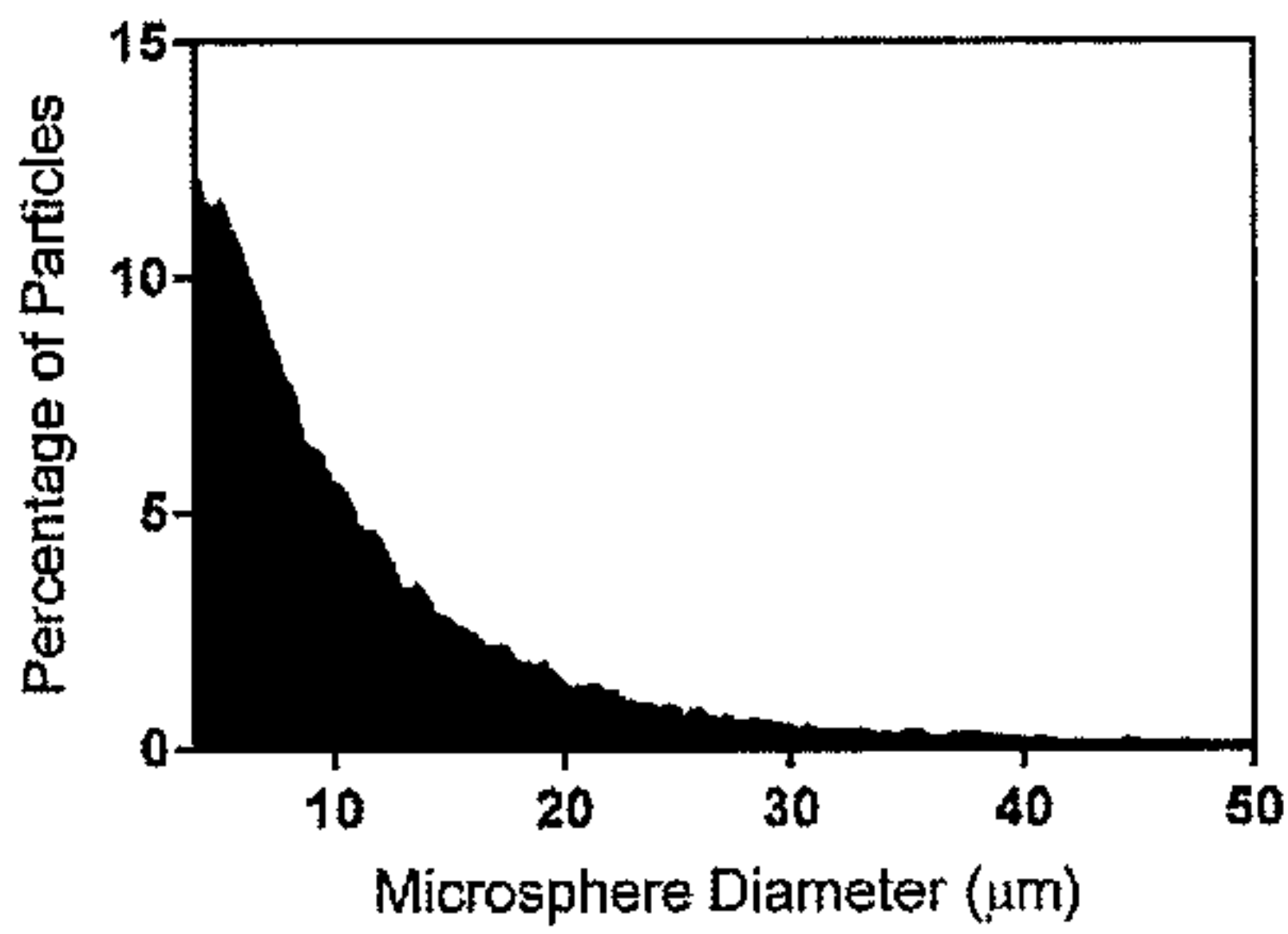


FIG. 20B

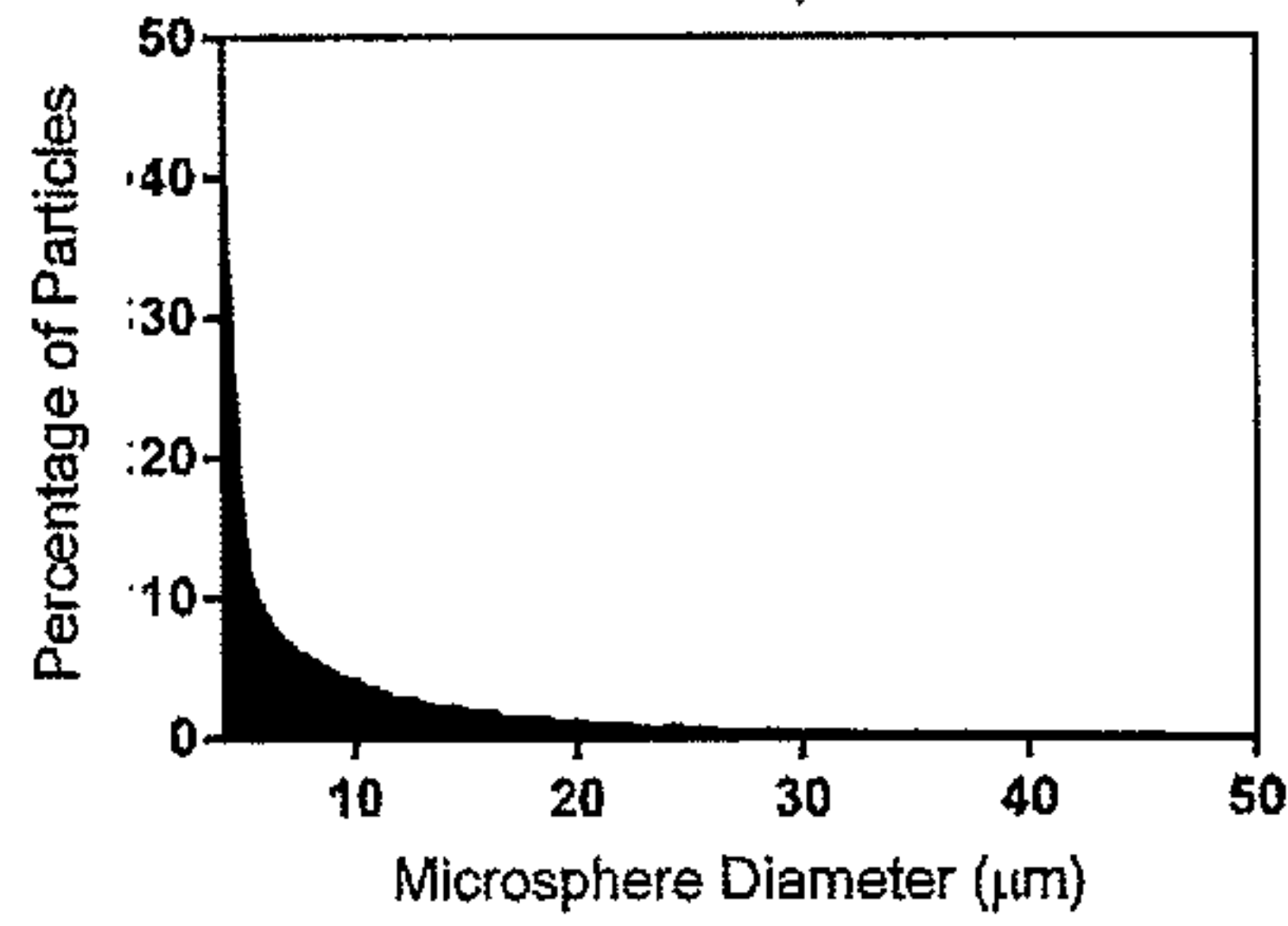


FIG. 20C

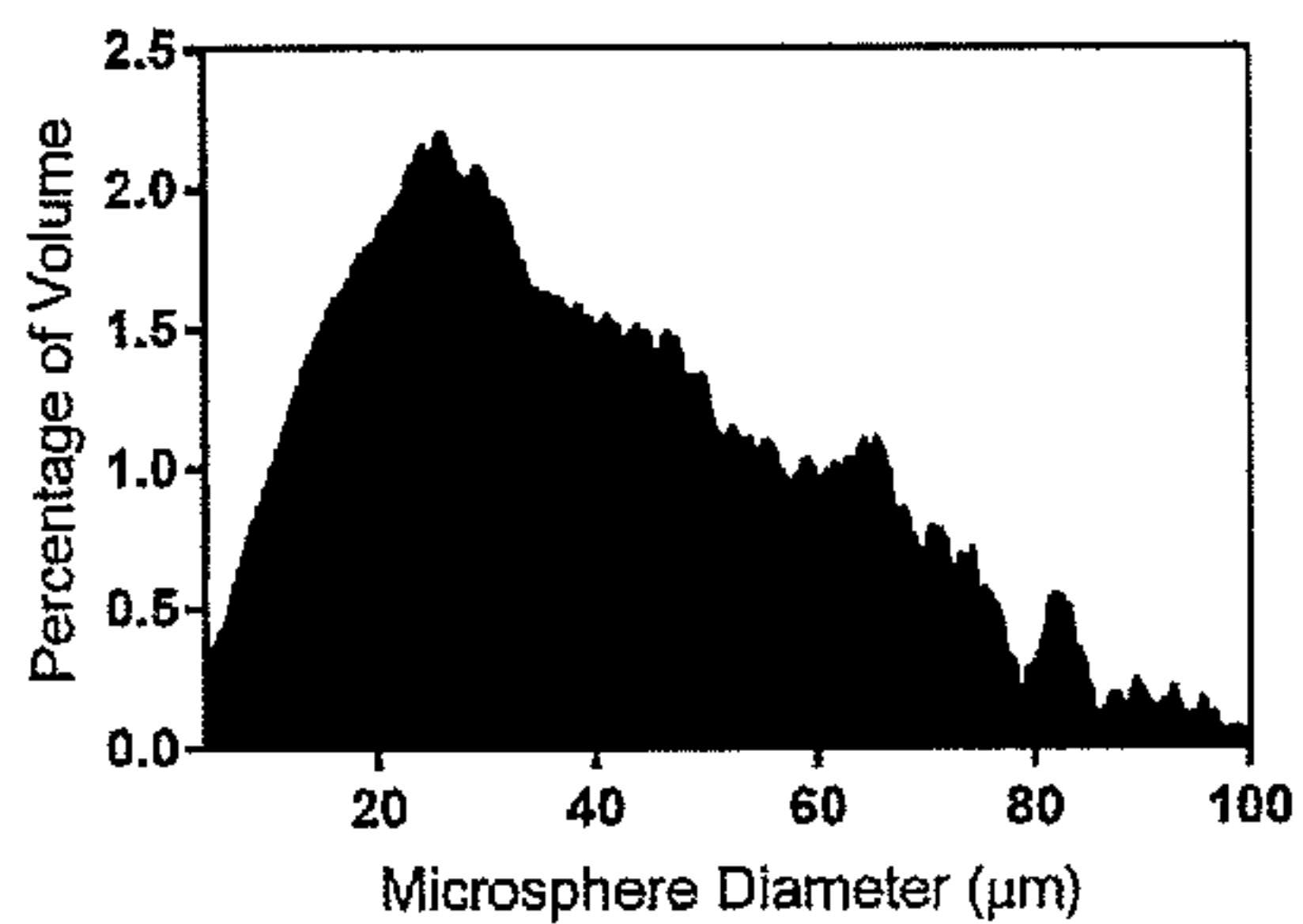


FIG. 20D

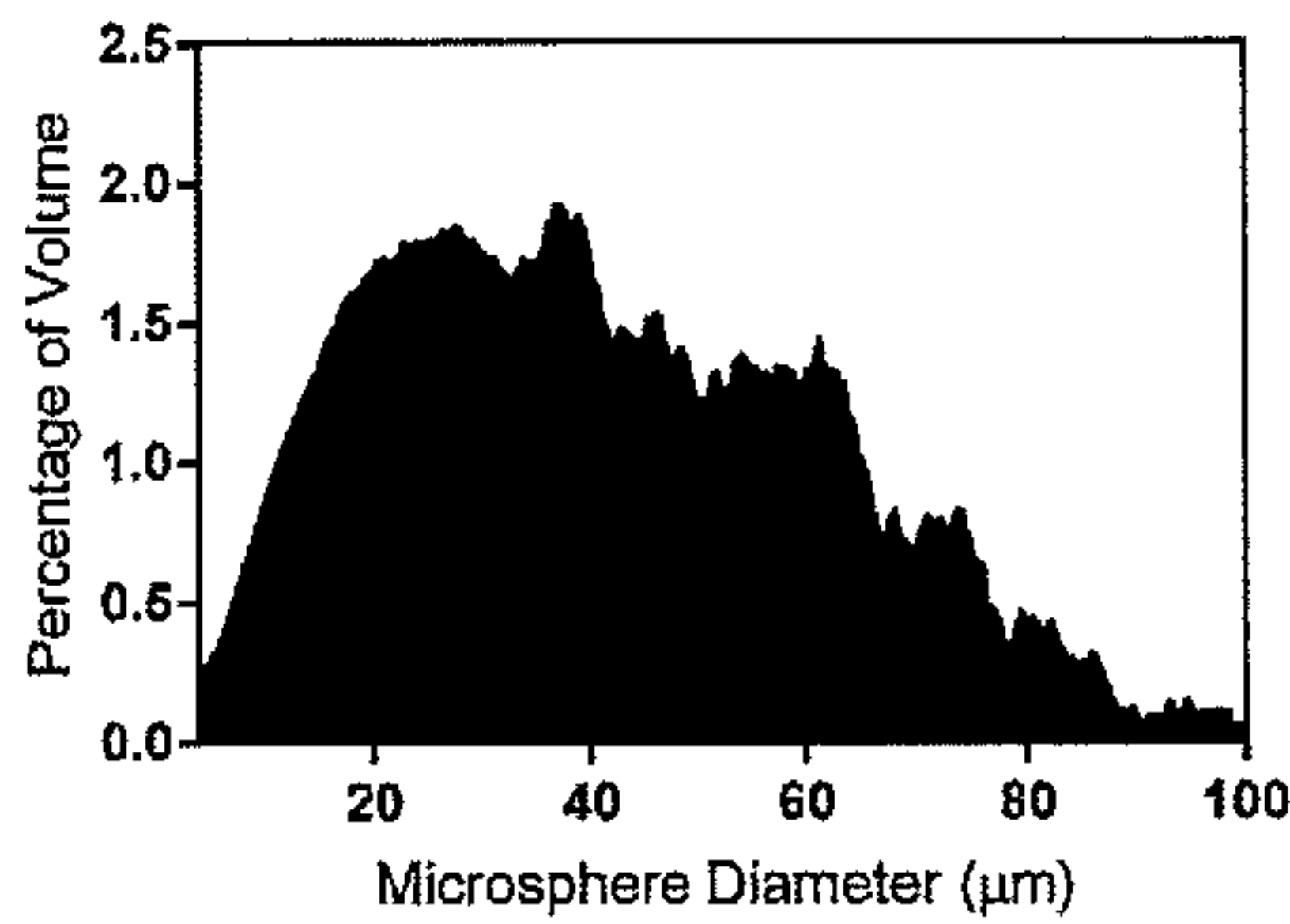


FIG. 20E

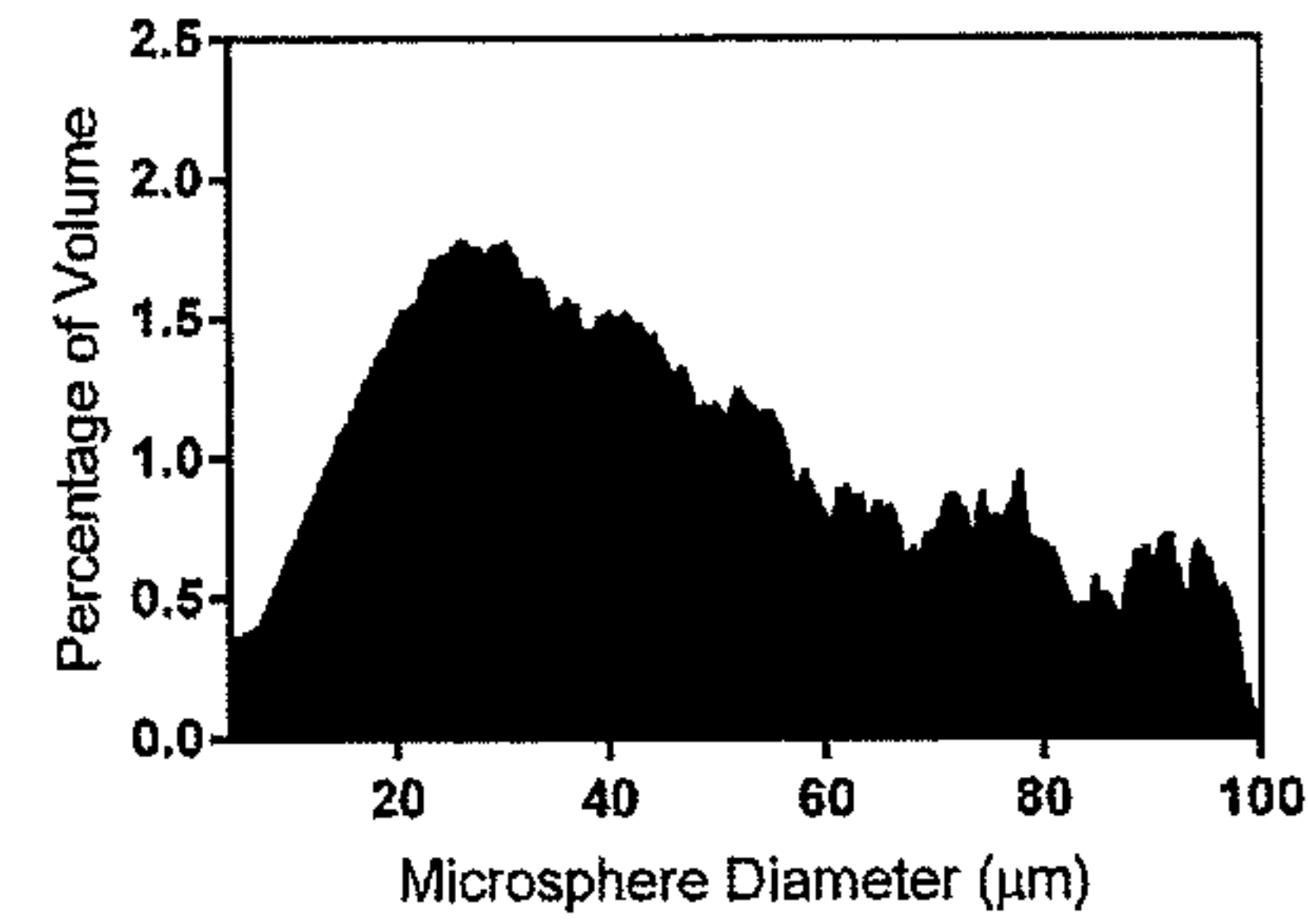


FIG. 20F



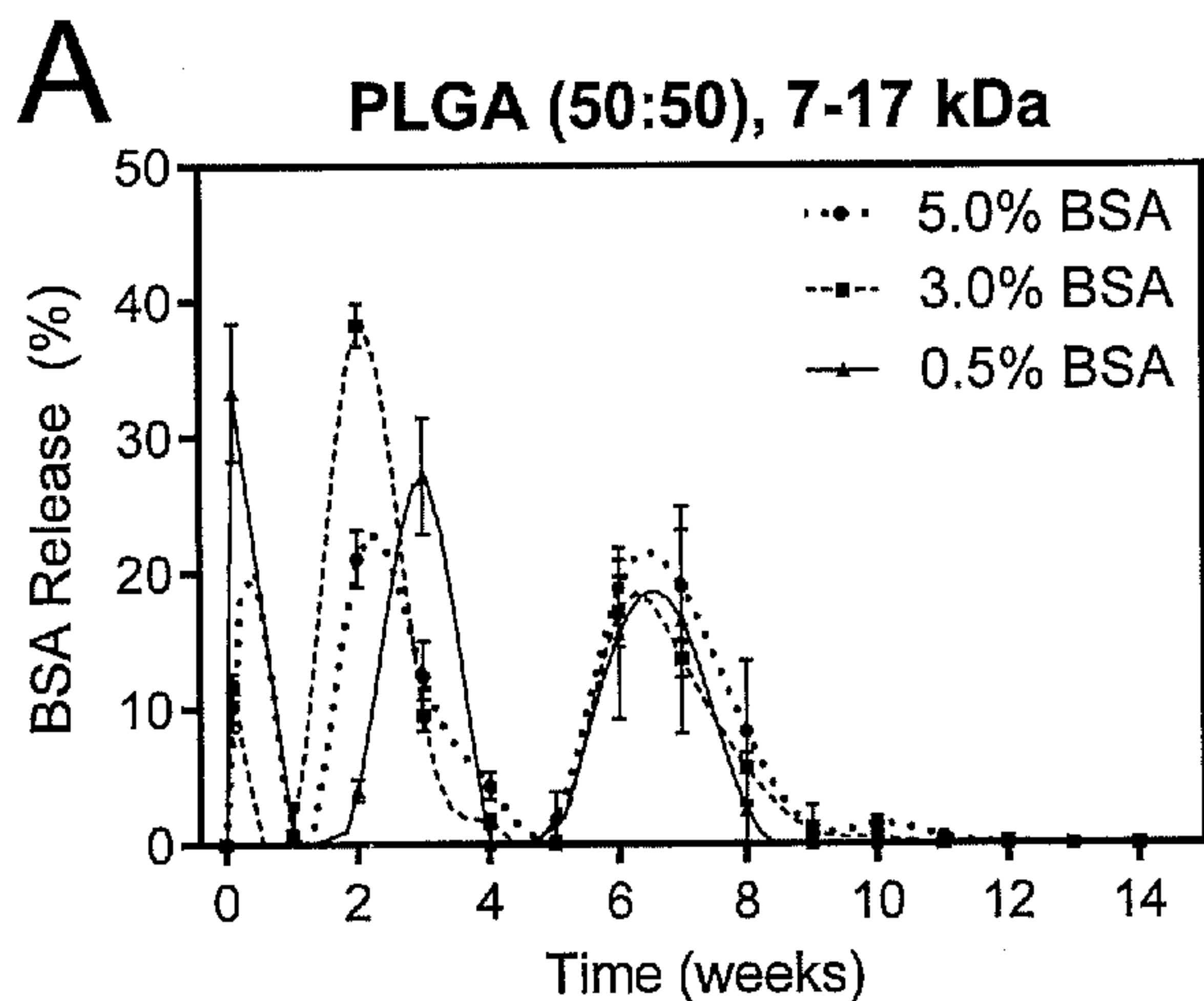


FIG. 21A

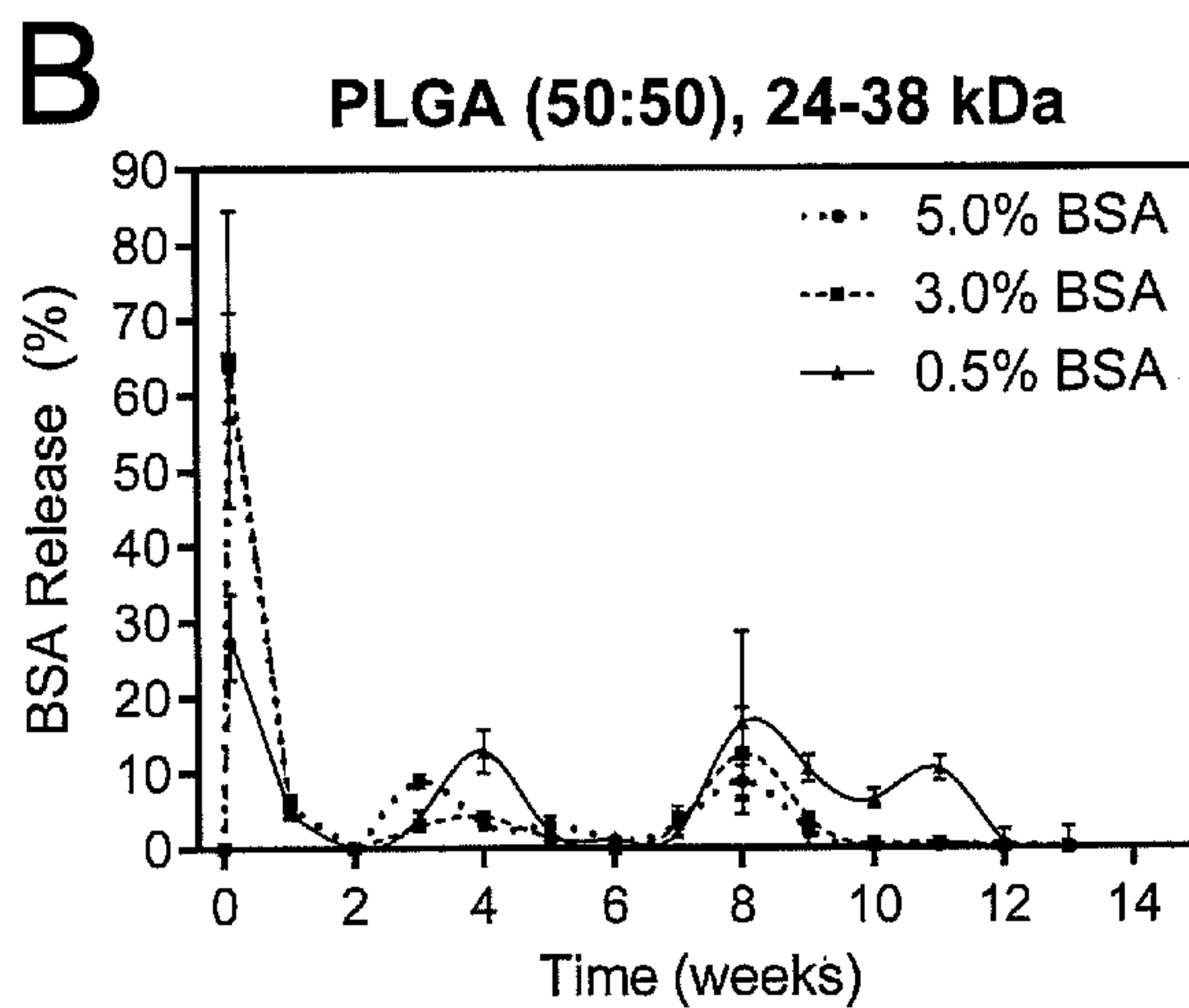


FIG. 21B

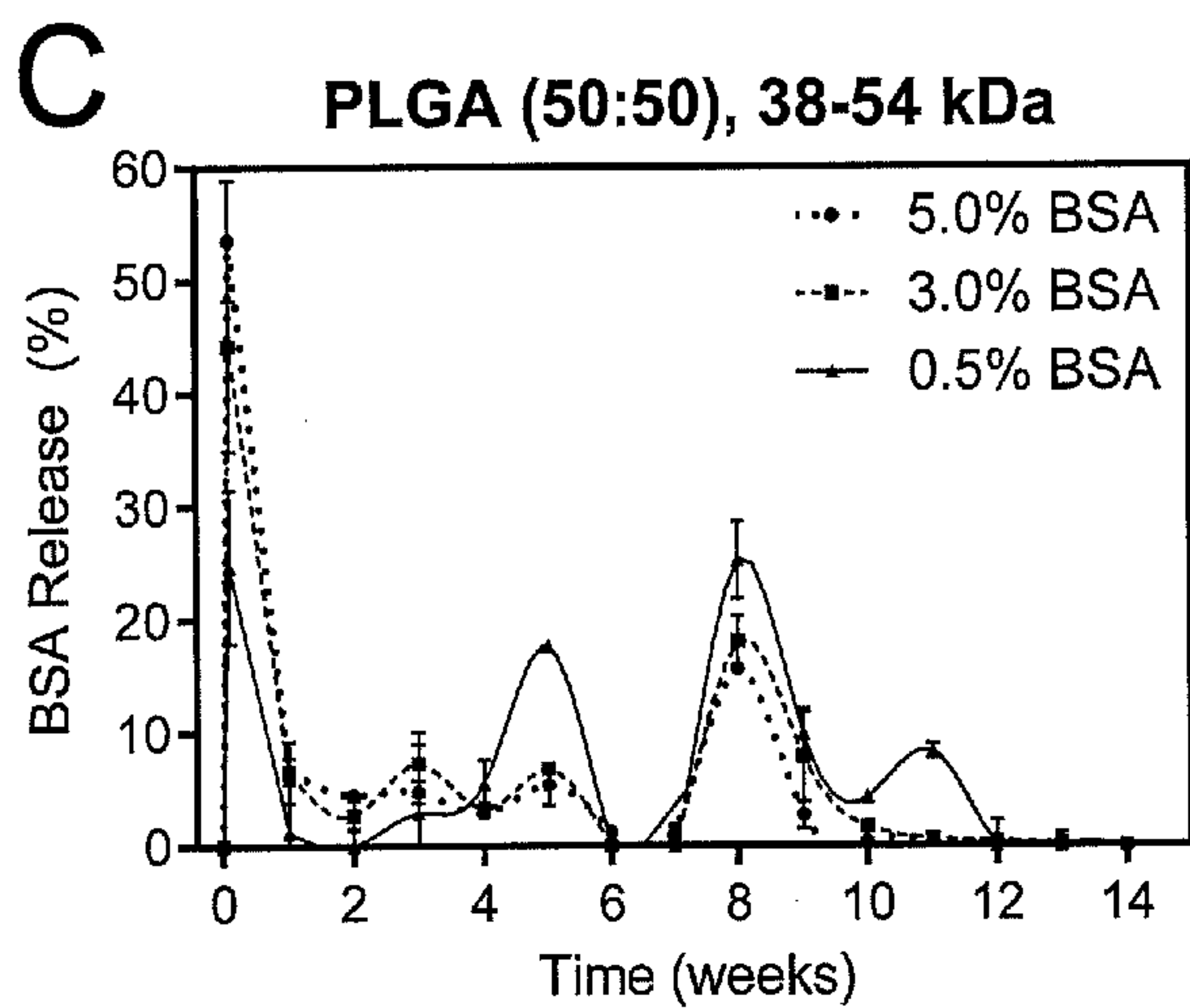


FIG. 21C

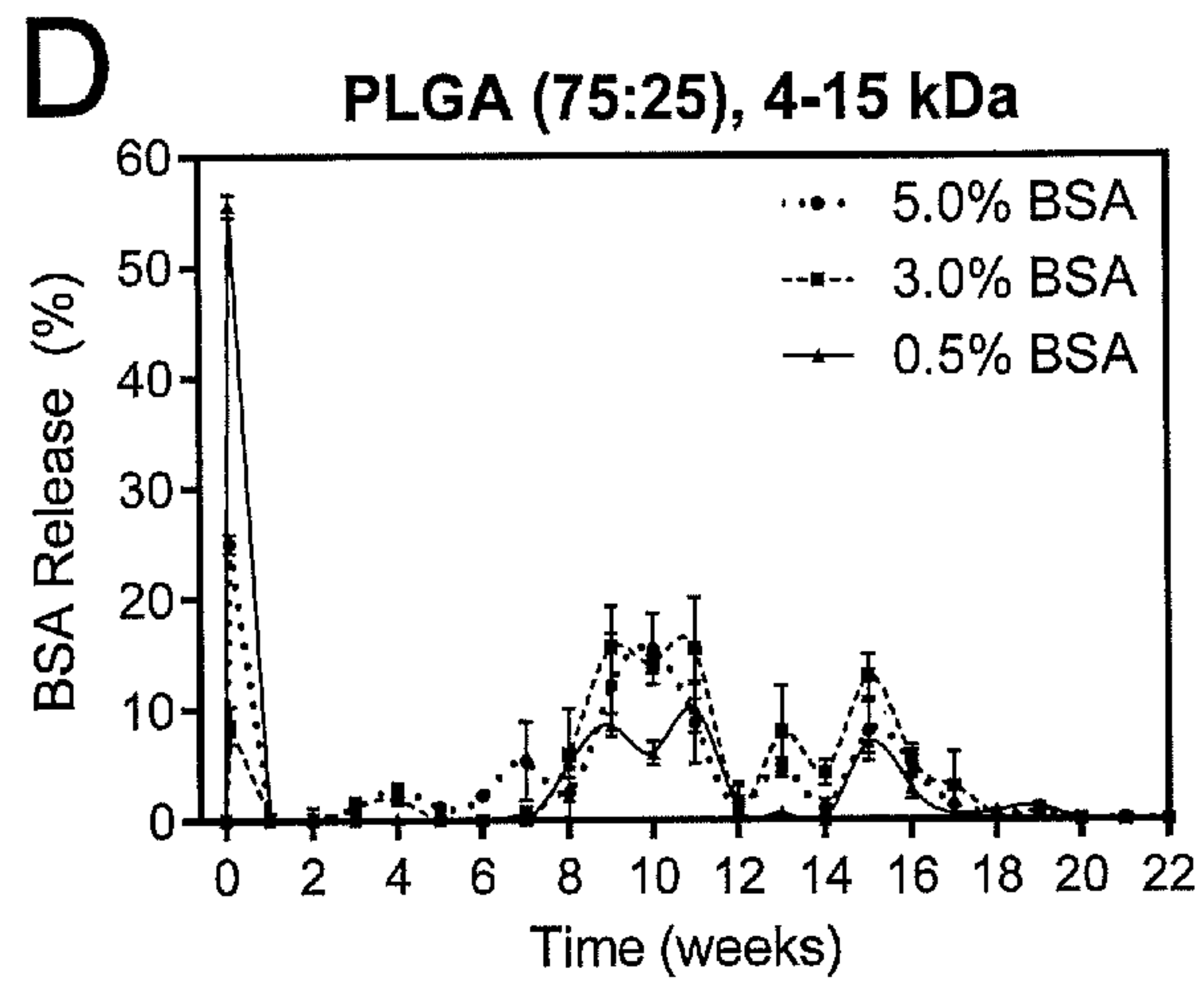


FIG. 21D

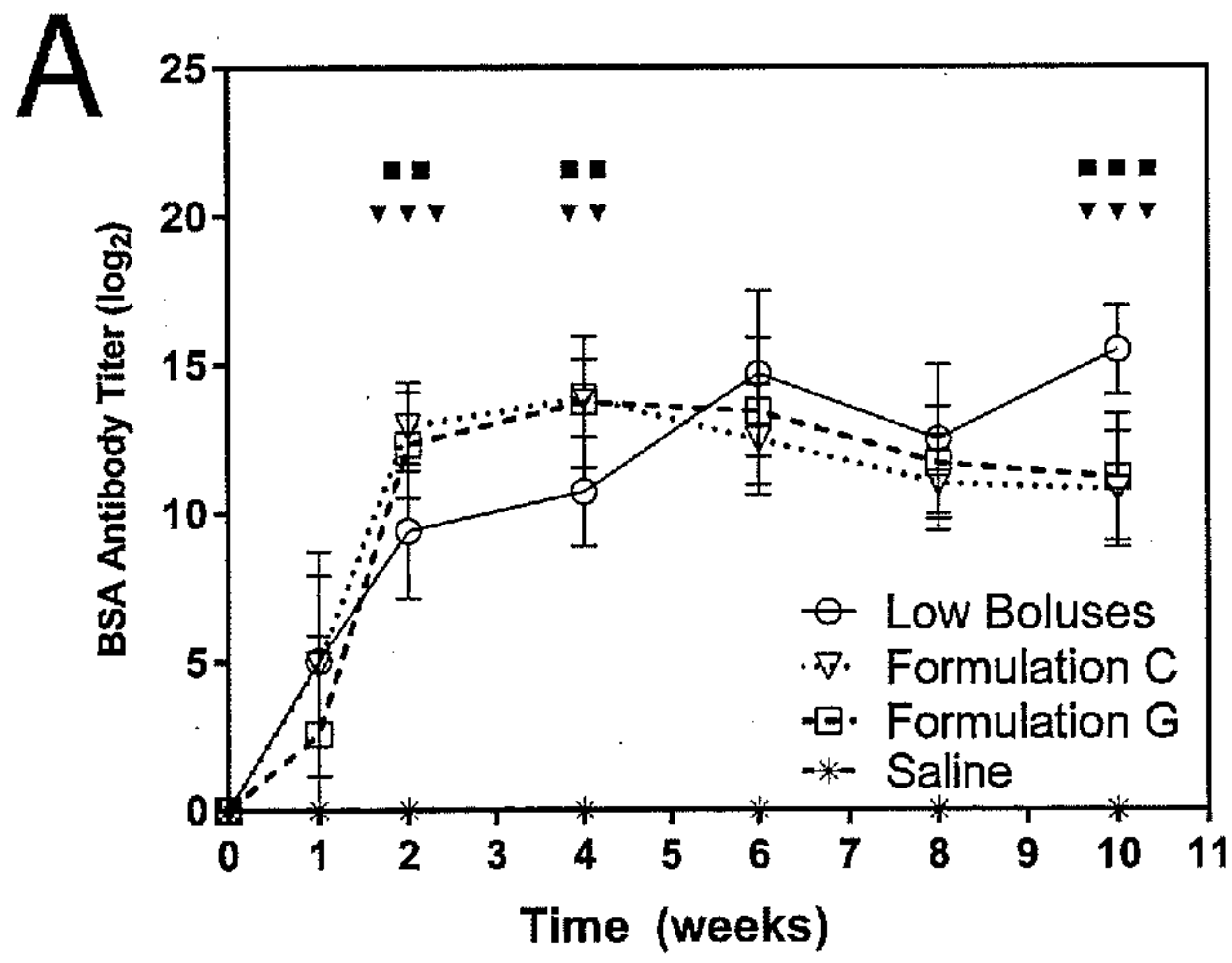


FIG. 22A

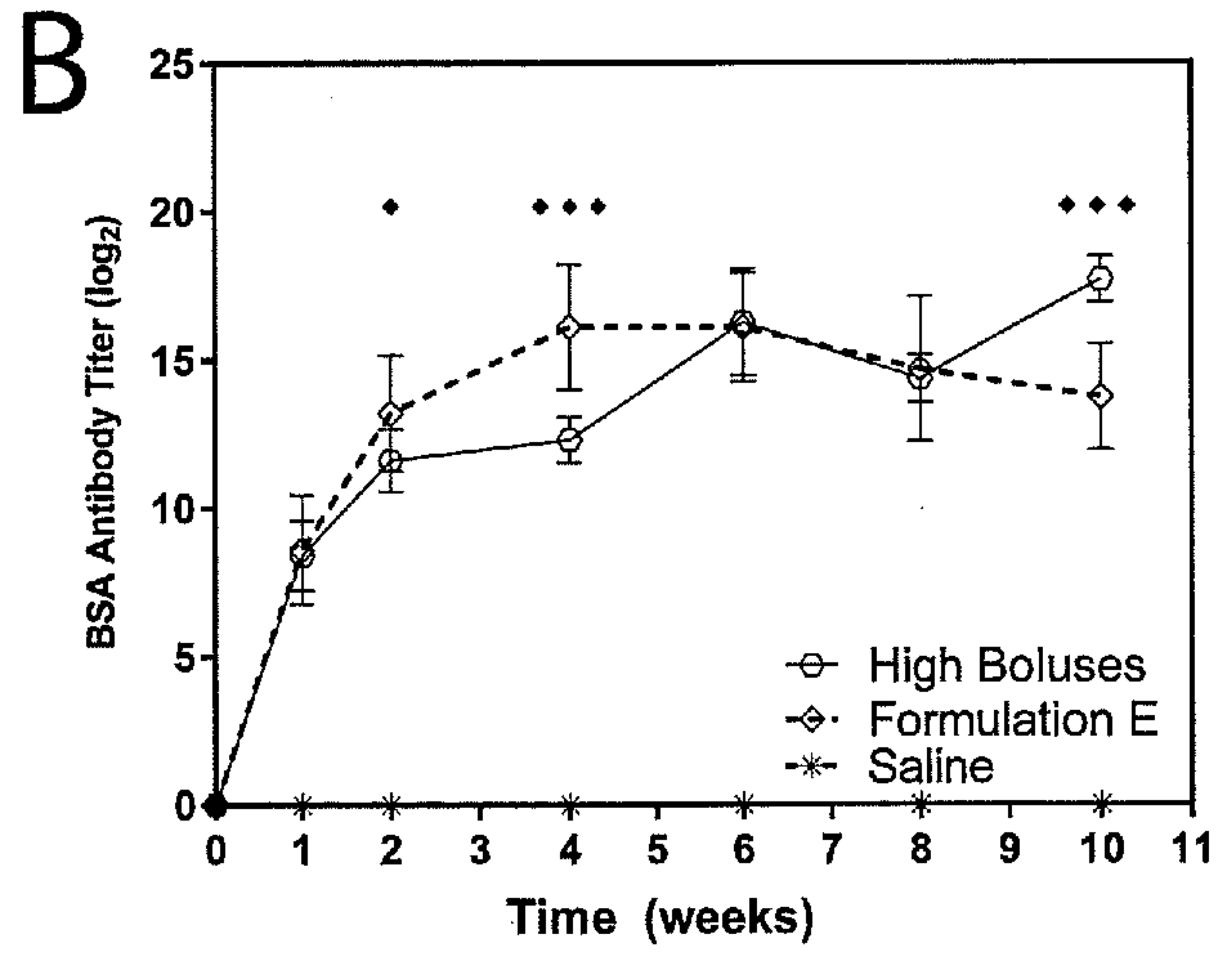


FIG. 22B

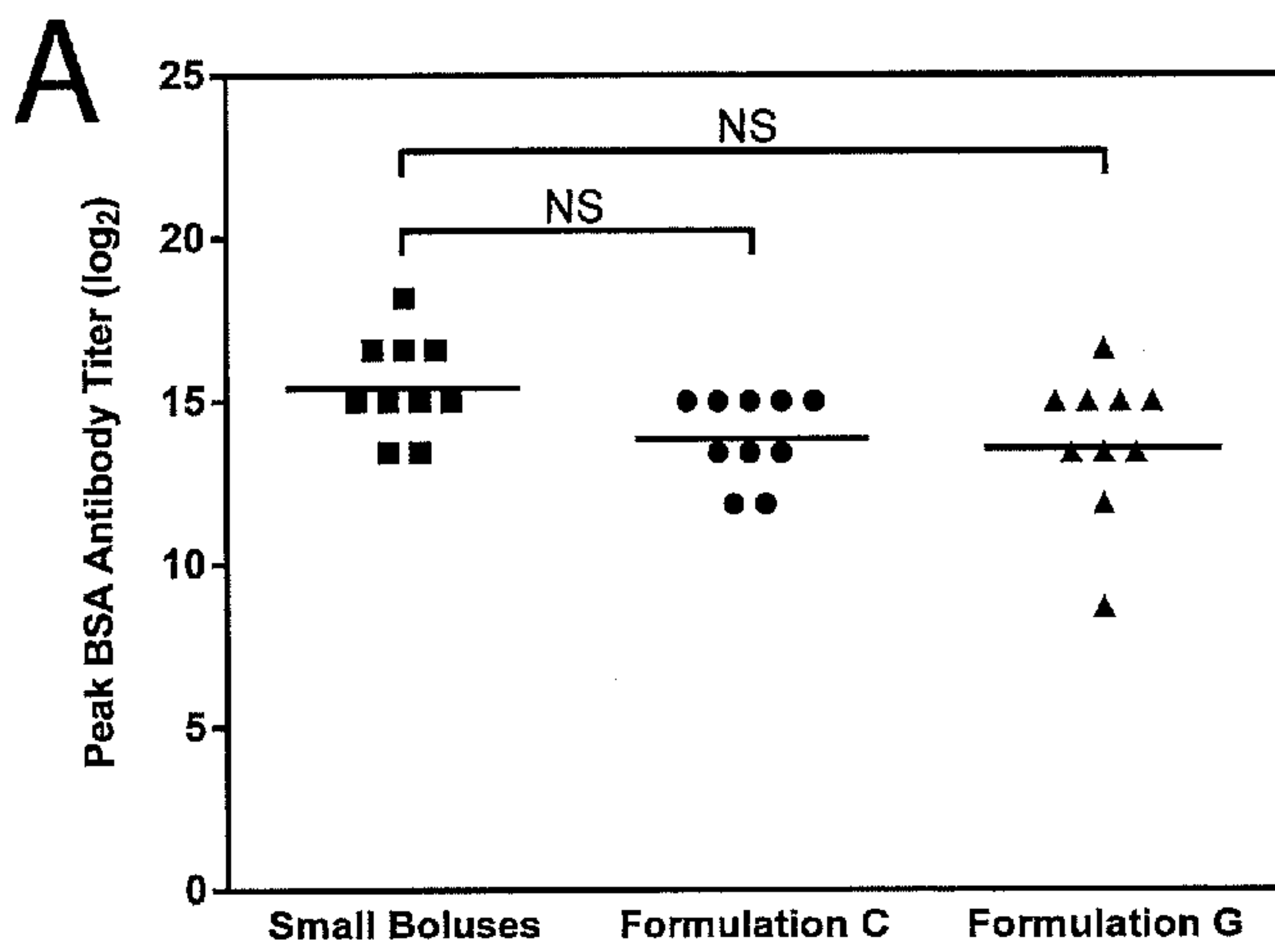


FIG. 23A

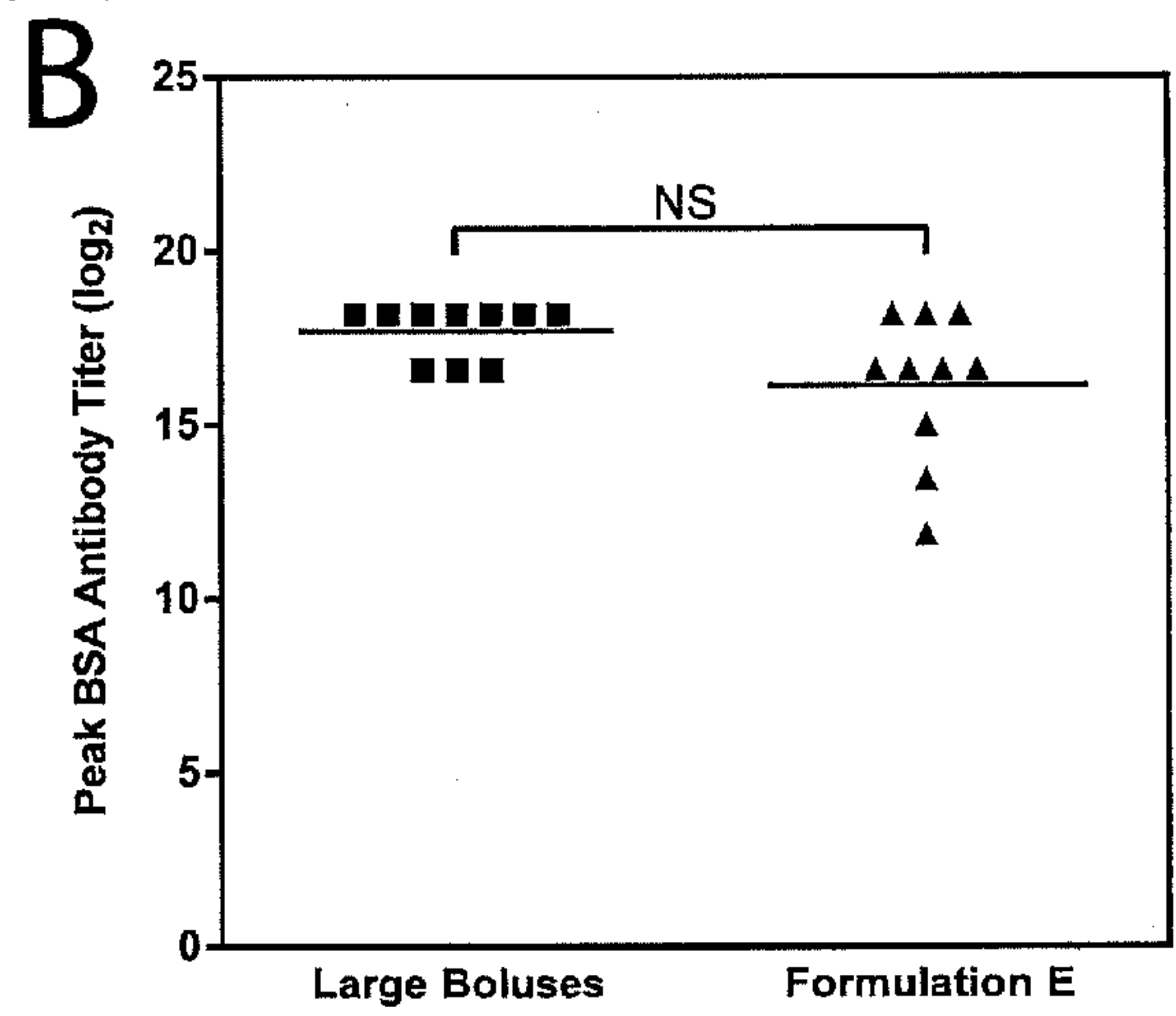


FIG. 23B

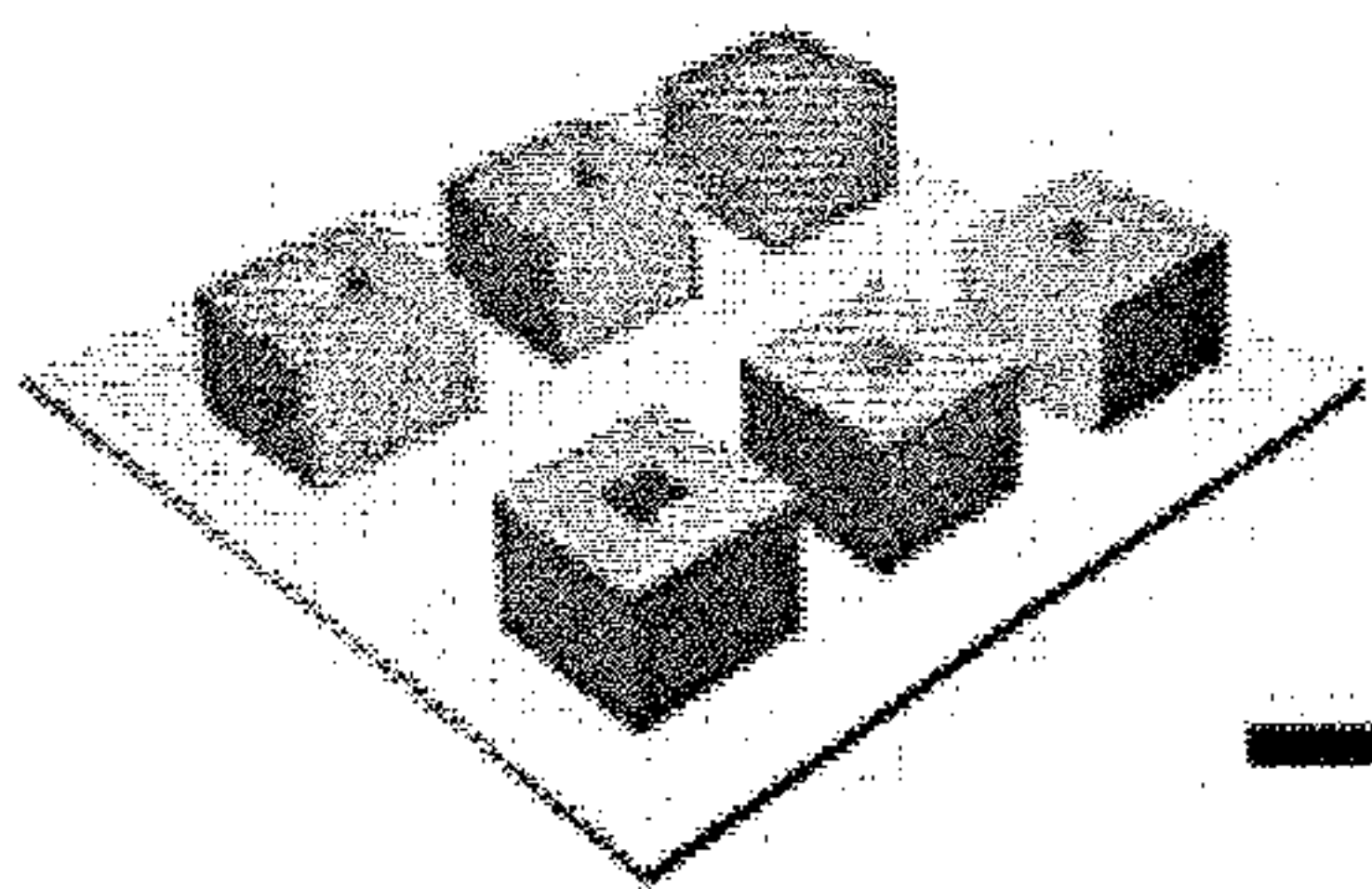


FIG. 24A

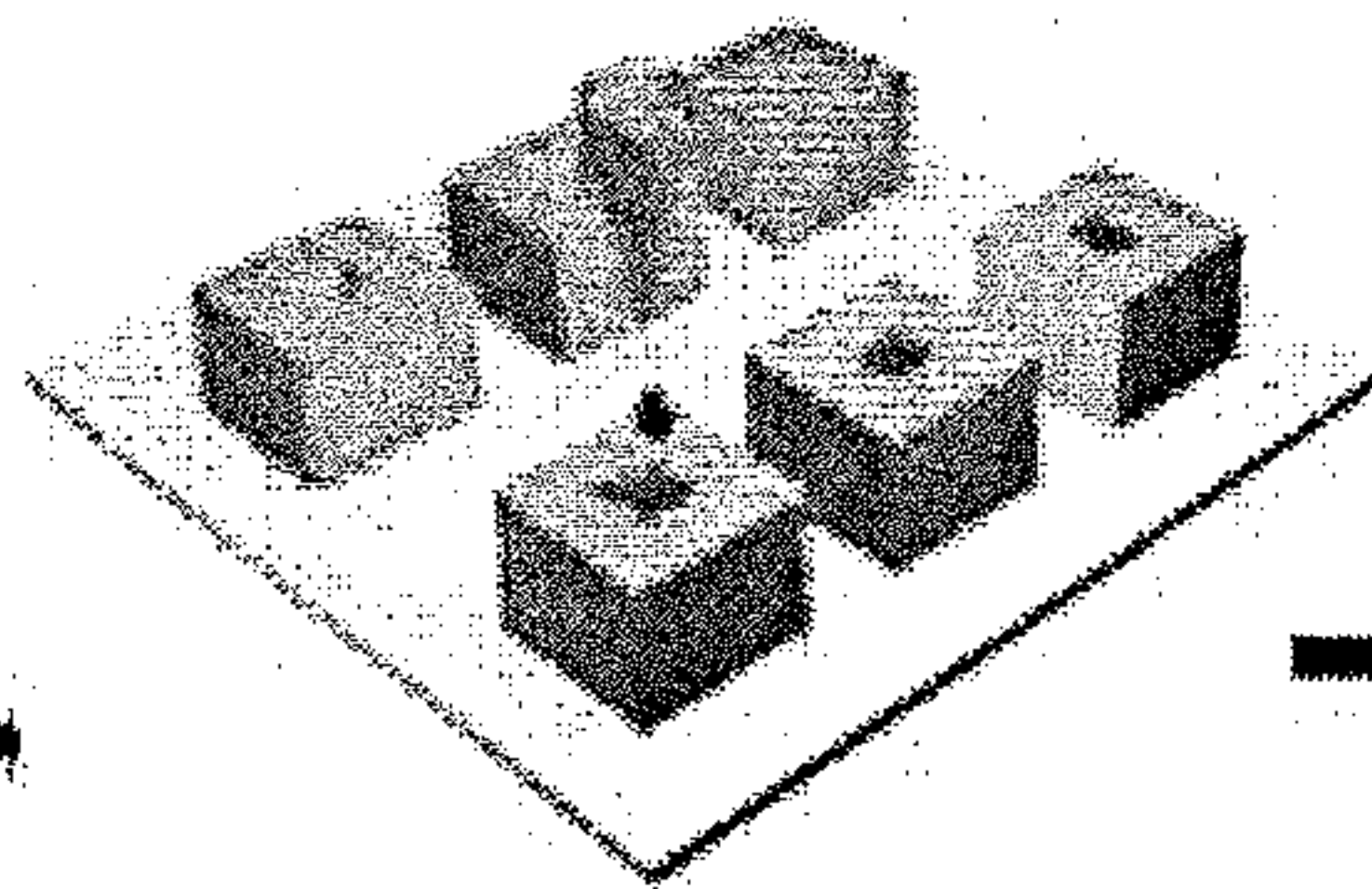


FIG. 24B

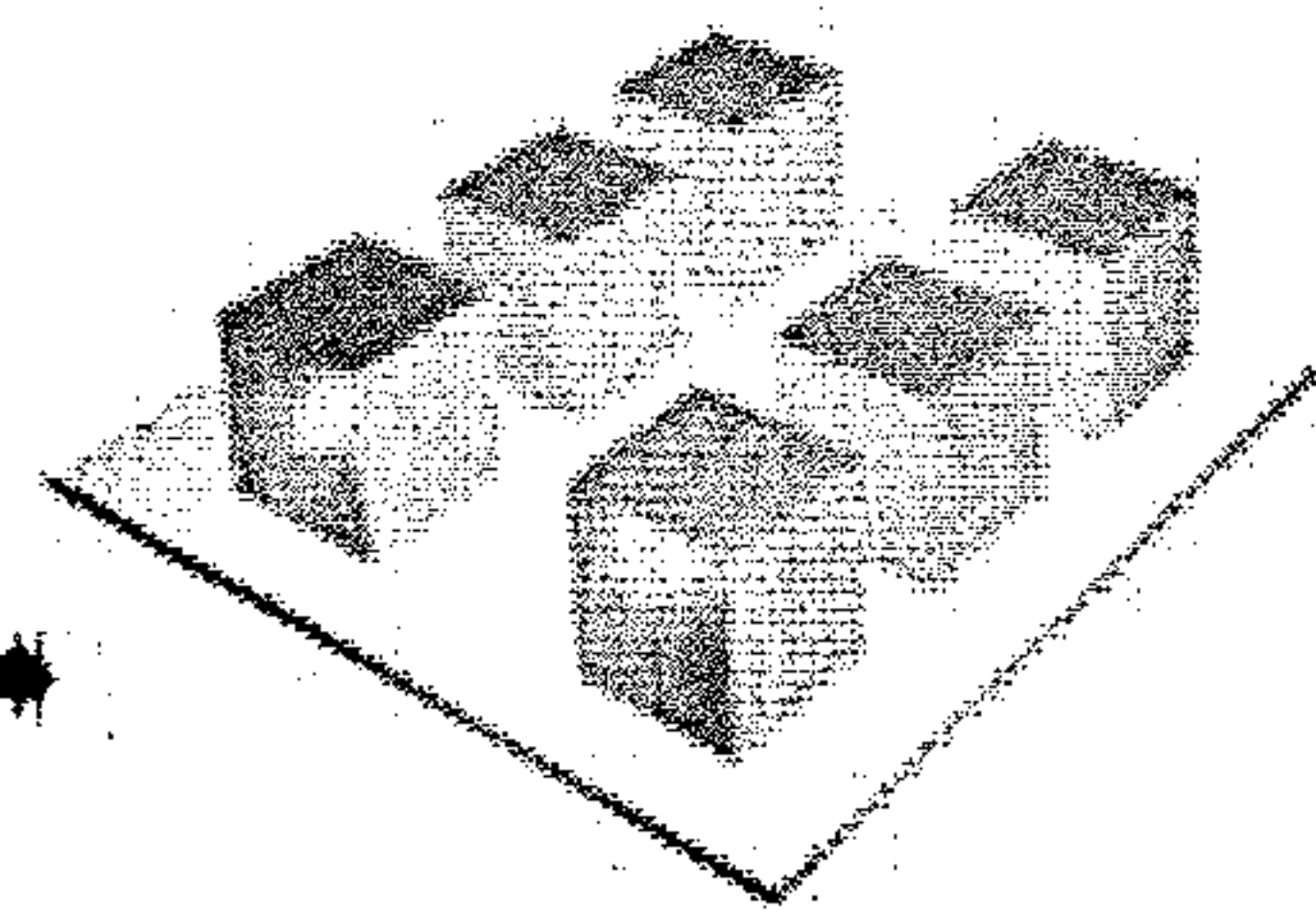


FIG. 24C

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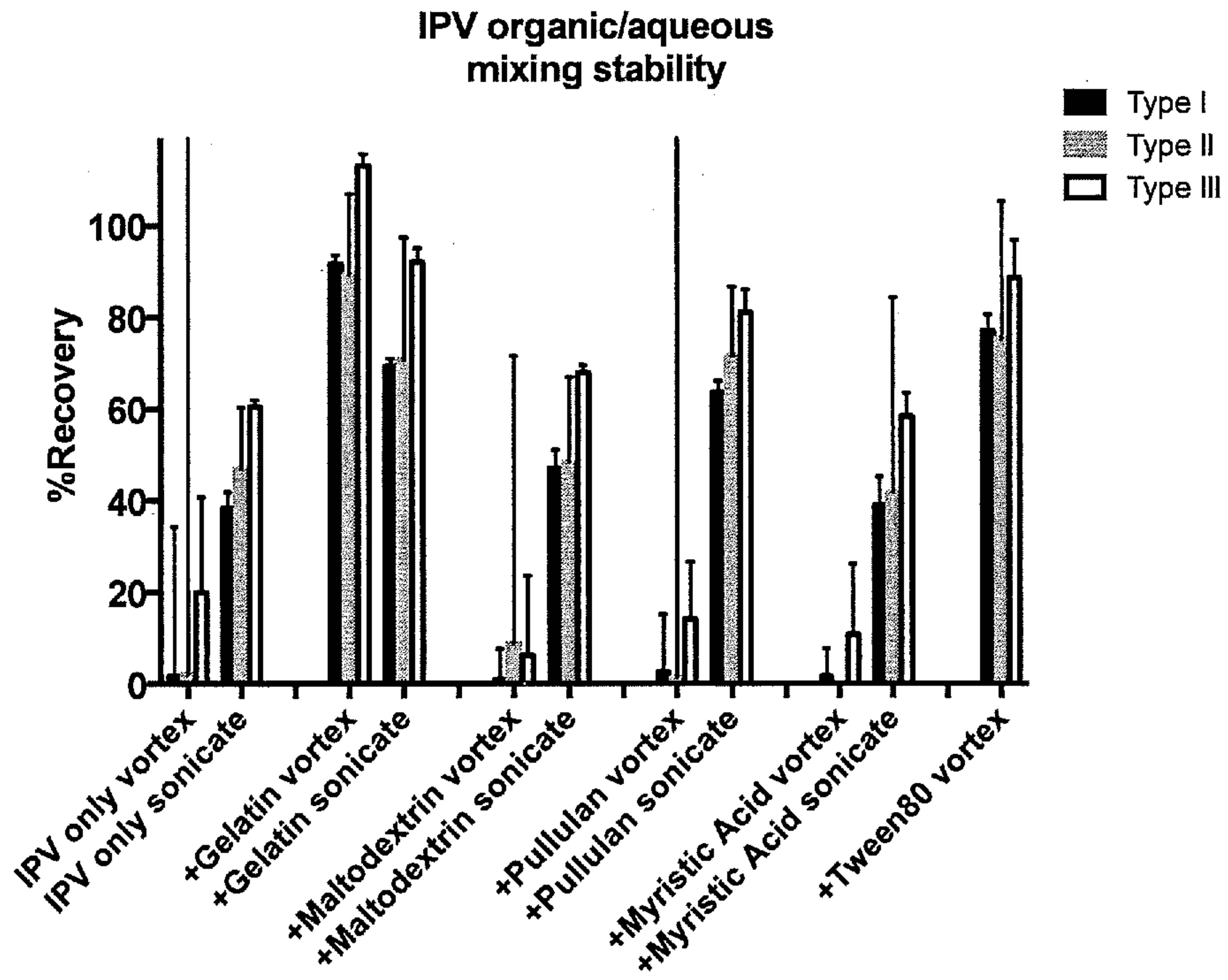


FIG. 25

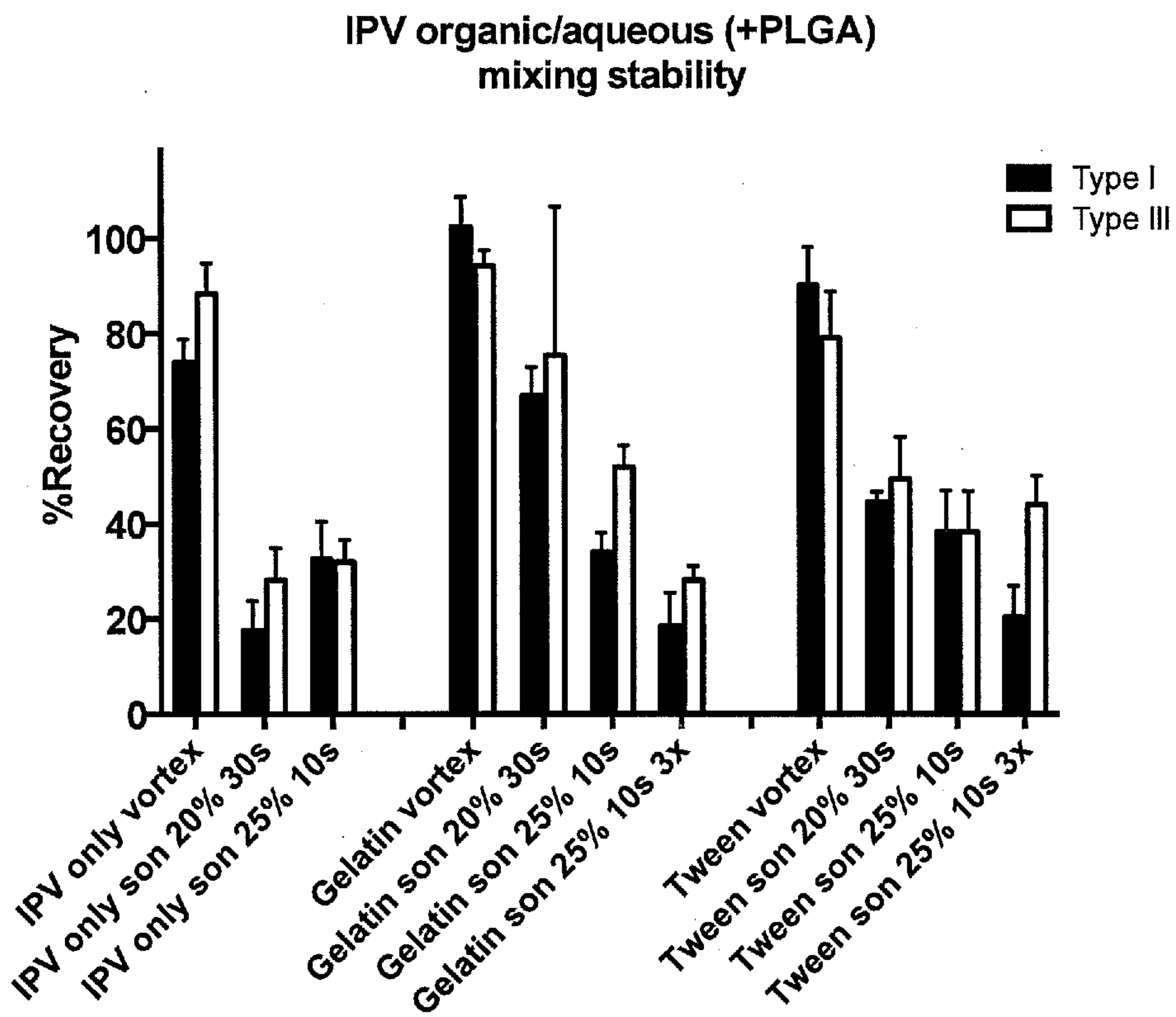


FIG. 26

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Drying excipients and method

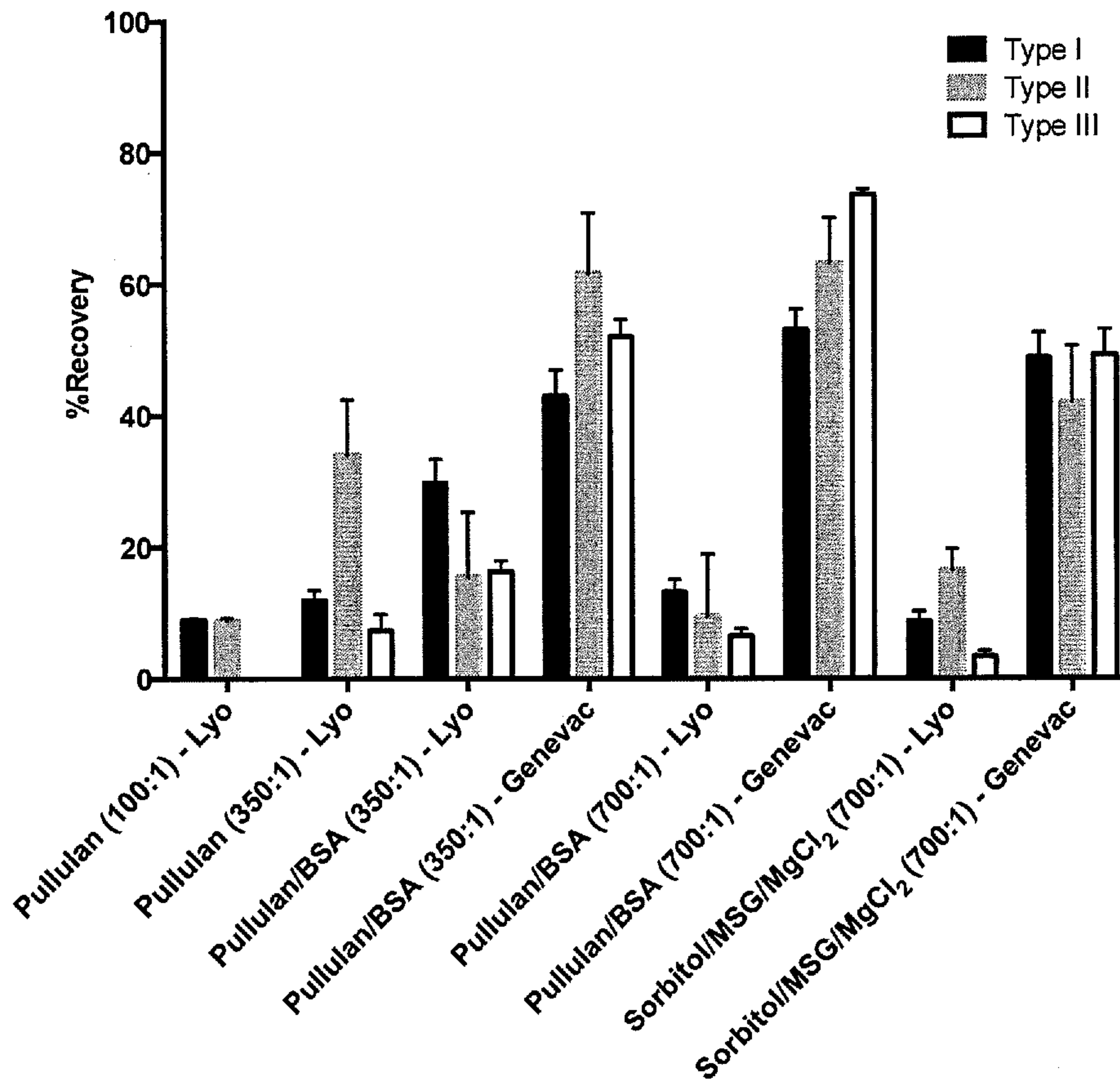


FIG. 27

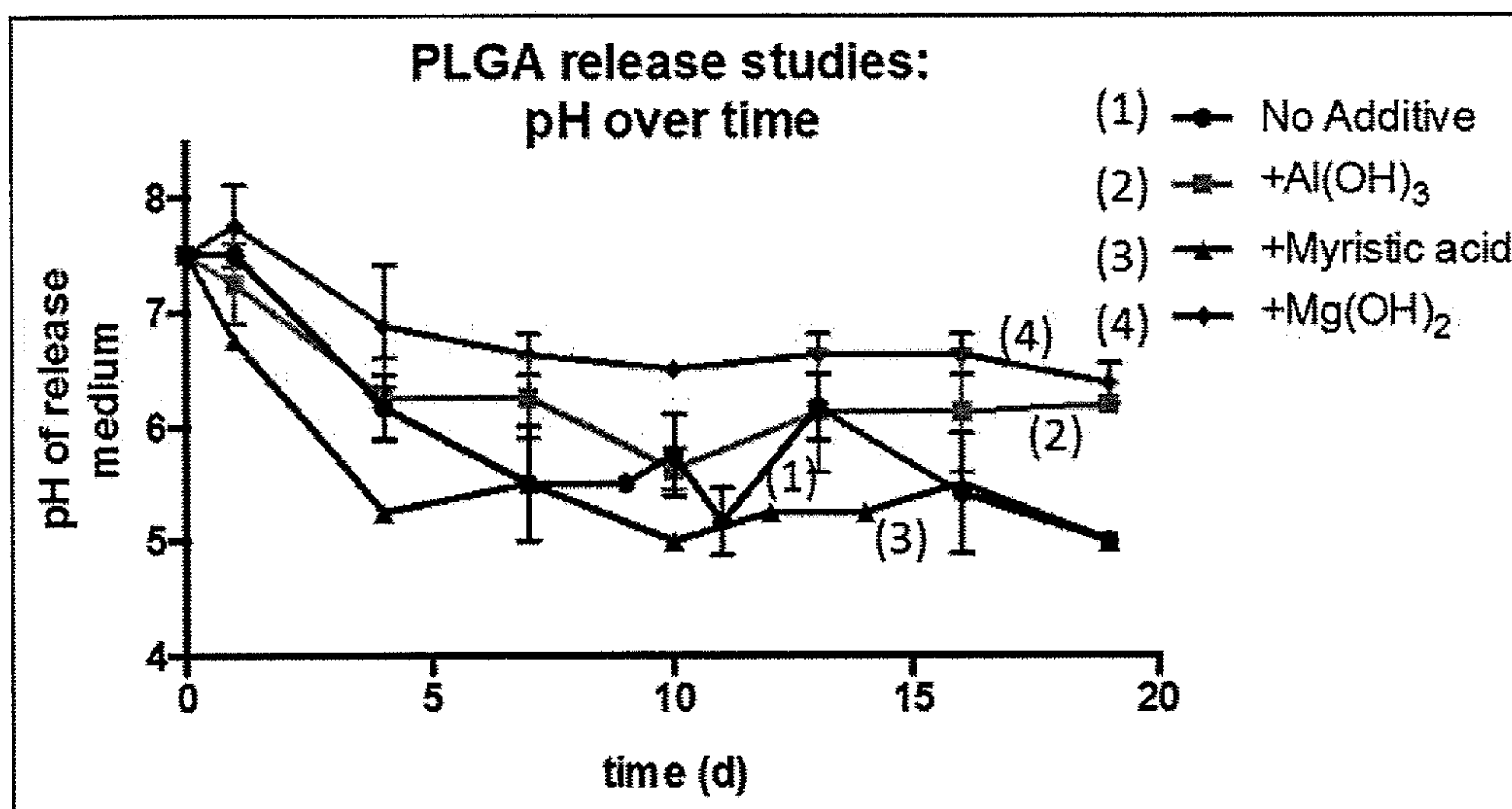


FIG. 28

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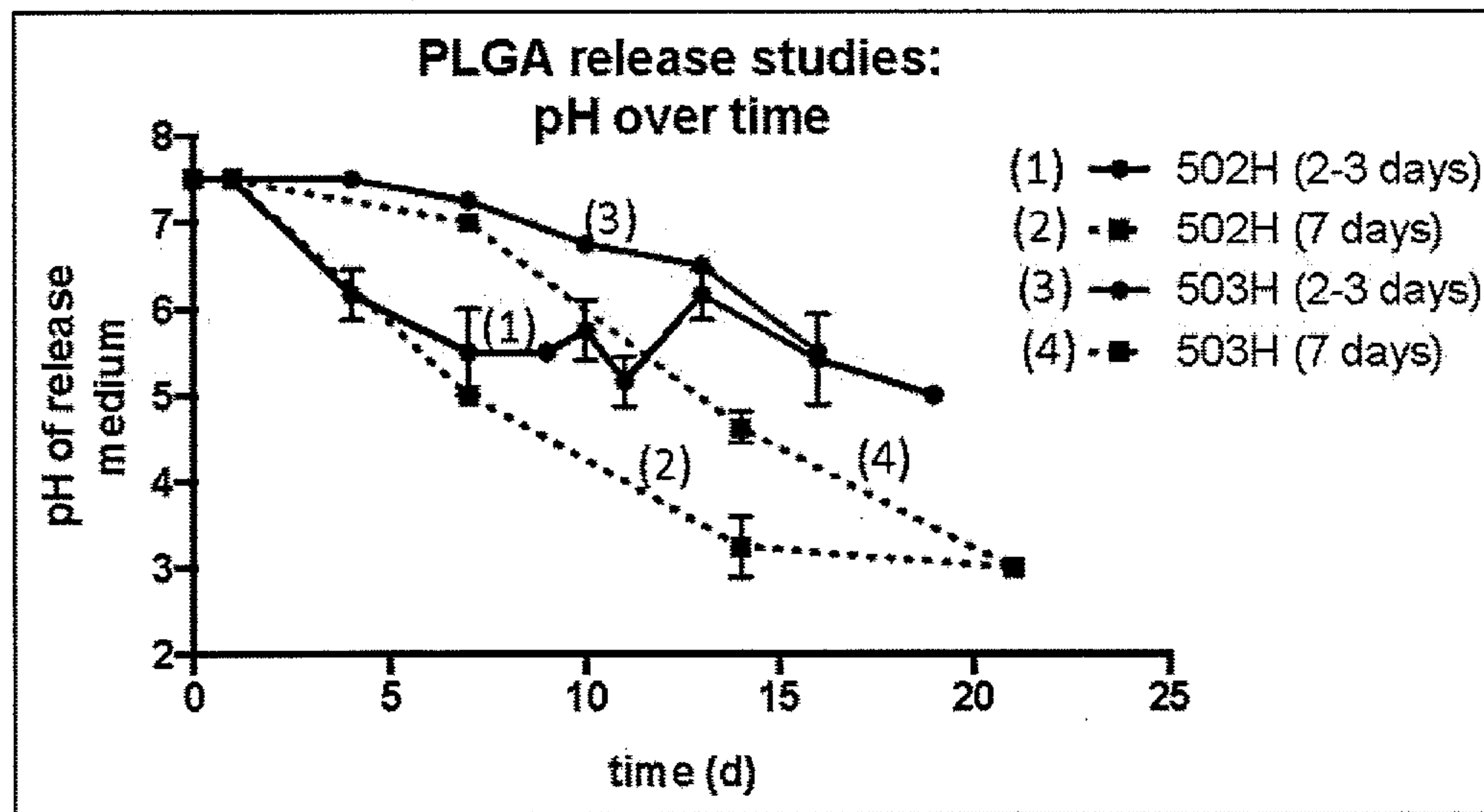
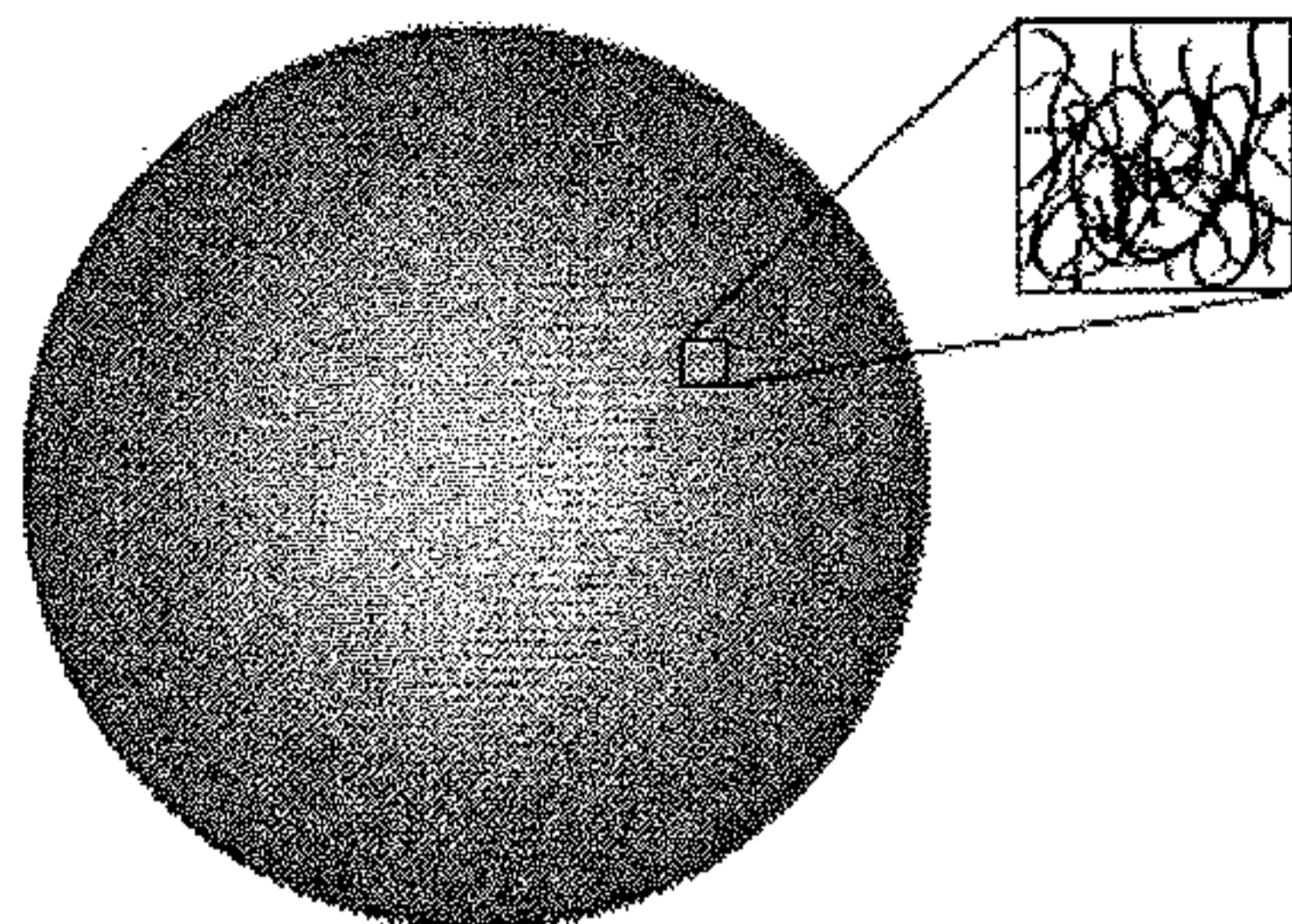
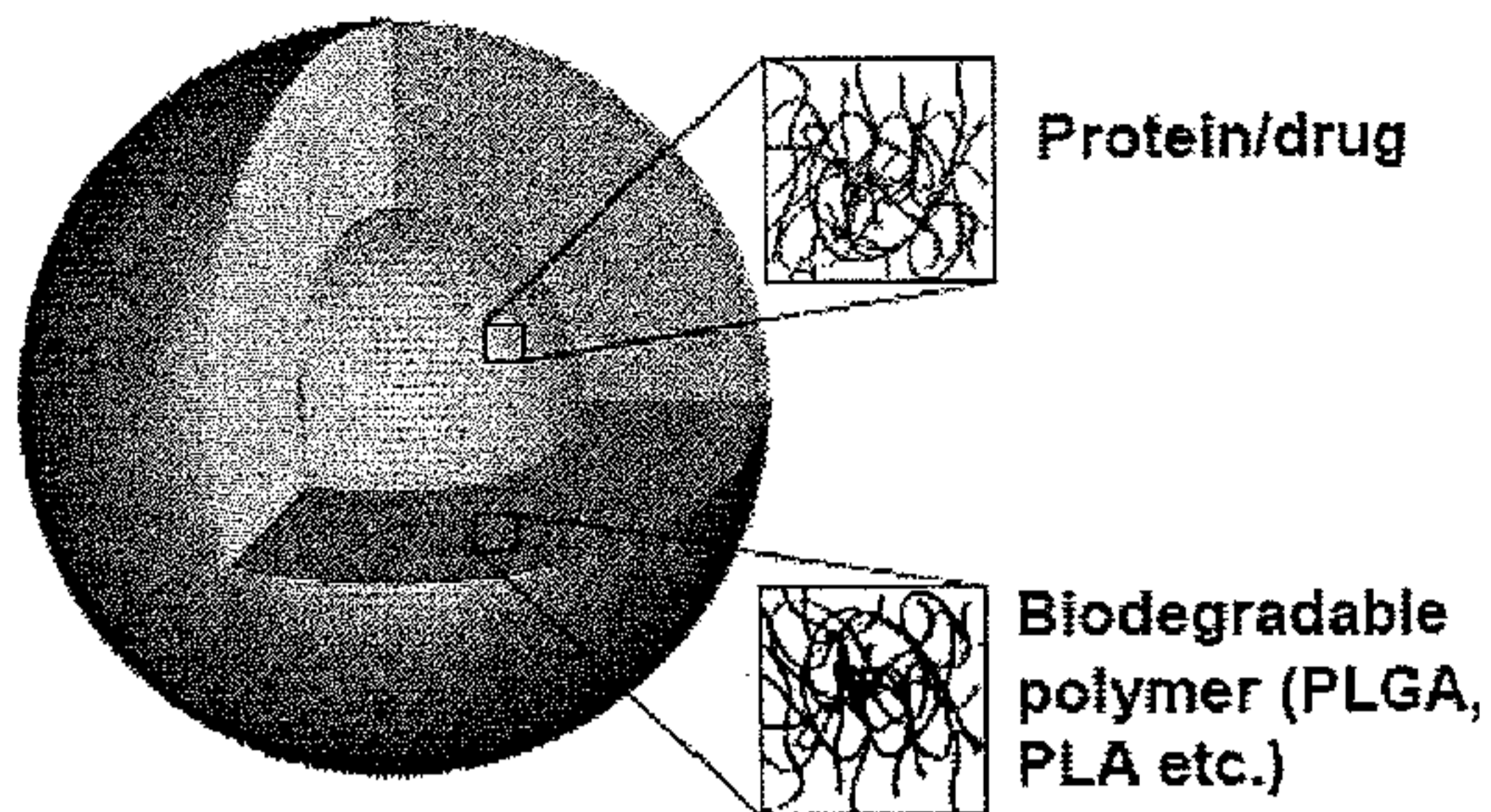


FIG. 29



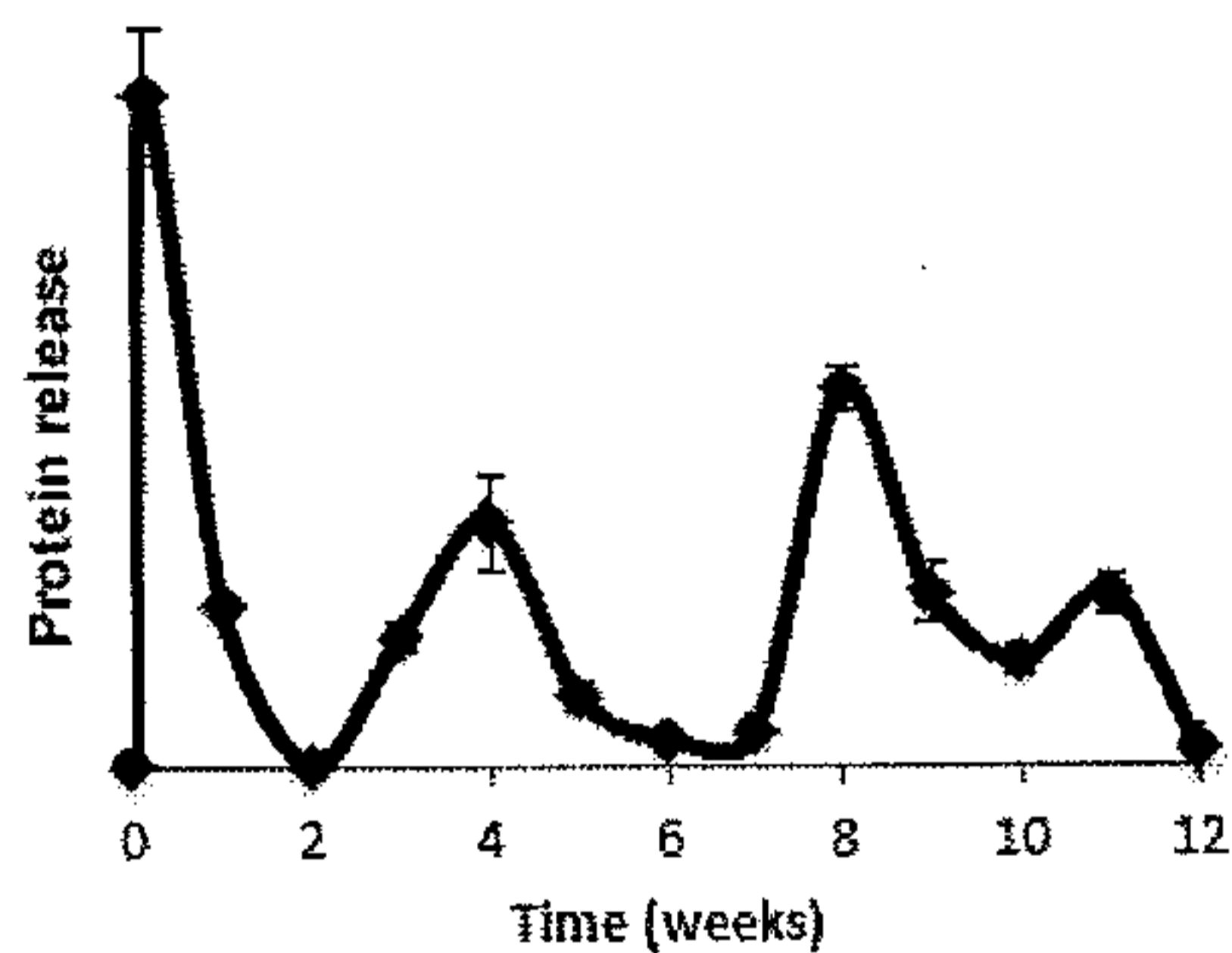
**Emulsion method**

**FIG. 11A**

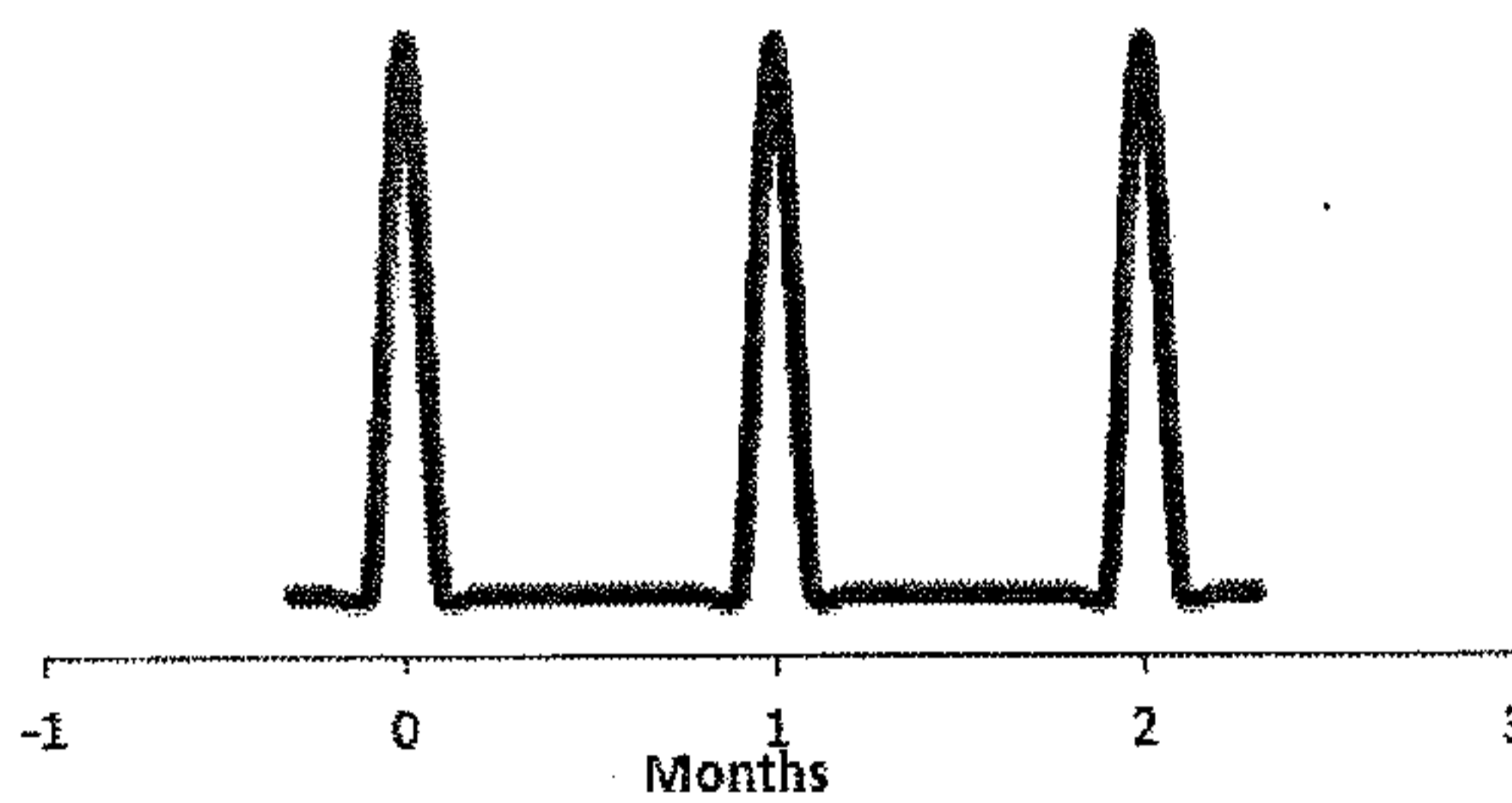


**3D printing**

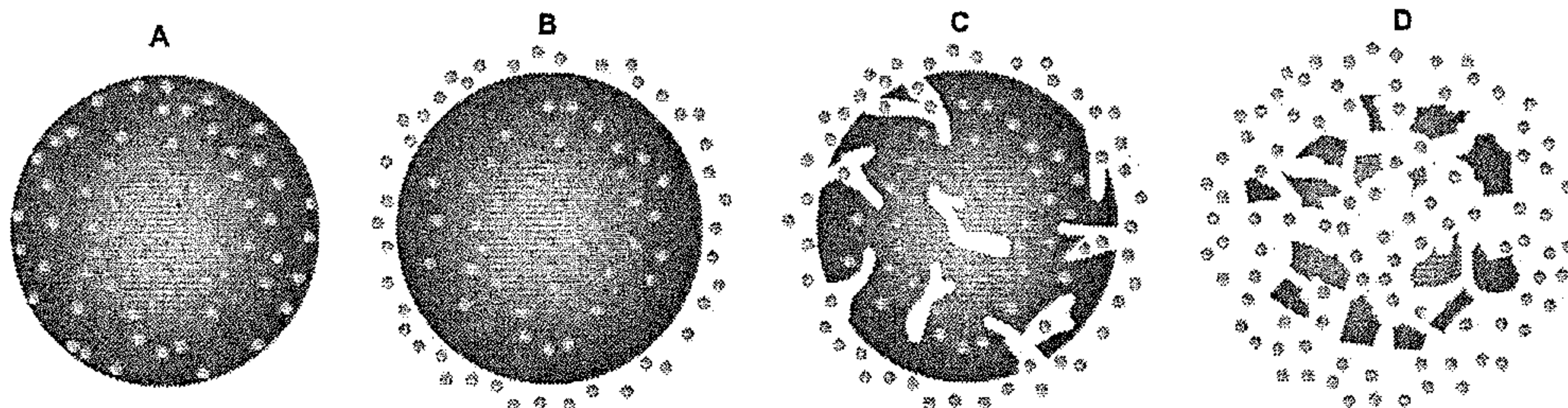
**FIG. 11B**



**FIG. 11C**



**FIG. 11D**



**FIG. 11E**