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Virus-Inspired Membrane Encapsulation of DNA Nanostructures To Achieve In Vivo Stability

Steven D. Perrault†,‡ and William M. Shih†,‡,§,*

†Wyss Institute for Biologically Inspired Engineering and ‡Biological Chemistry and Molecular Pharmacology, Harvard Medical School, Boston, Massachusetts 02115, United States, and §Department of Cancer Biology, Dana Farber Cancer Institute, Boston, Massachusetts 02115, United States

ABSTRACT DNA nanotechnology enables engineering of molecular-scale devices with exquisite control over geometry and site-specific functionalization. This capability promises compelling advantages in advancing nanomedicine; nevertheless, instability in biological environments and innate immune activation remain as obstacles for in vivo application. Natural particle systems (i.e., viruses) have evolved mechanisms to maintain structural integrity and avoid immune recognition during infection, including encapsulation of their genome and protein capsid shell in a lipid envelope. Here we introduce virus-inspired enveloped DNA nanostructures as a design strategy for biomedical applications. Achieving a high yield of tightly wrapped unilamellar nanostructures, mimicking the morphology of enveloped virus particles, required precise control over the density of attached lipid conjugates and was achieved at 1 per ~180 nm². Envelopment of DNA nanostructures in PEGylated lipid bilayers conferred protection against nuclease digestion. Immune activation was decreased 2 orders of magnitude below controls, and pharmacokinetic bioavailability improved by a factor of 17. By establishing a design strategy suitable for biomedical applications, we have provided a platform for the engineering of sophisticated, translation-ready DNA nanodevices.

KEYWORDS: DNA · nanotechnology · lipid bilayer · in vivo · nanostructure · imaging · immune · pharmacokinetics · biodistribution · PEG

Biomedical nanotechnology has undergone considerable progress over the past five decades, from an initial demonstration of liposomes1,2 to that of a logic-gated nanorobot.3 Much of this development has focused on improving the detection and treatment of cancer by increasing the bioavailability and targeting specificity of anticancer agents.4–11 The rapidly growing field of structural DNA nanotechnology12–16 could advance these aims by expanding the range of available nanoparticle geometries and improving the precision of ligand functionalization, two key design parameters. Even more exciting is DNA nanotechnology’s potential for engineering of highly sophisticated nanoscale devices. Recent studies include the nanorobot,3 programmable immunoadjuvants,17,18 a synthetic membrane channel,19 and a molecular cascade capable of autonomously processing multiple inputs to determine cell phenotype.20 Translation of such devices into biomedical applications requires molecular engineers to first address the susceptibility of DNA nanostructures to nuclease degradation,21 as well as their ability to activate an inflammatory immune response.17,22 Inspired by the stability provided to enveloped viruses via their lipid membranes, we developed enveloped DNA nanostructures that address the above challenges (Figure 1a).

RESULTS AND DISCUSSION

To mimic the geometry of a viral protein capsid shell, we designed a wireframe DNA nano-octahedron (DNO) with an estimated diameter of ~50 nm (Supporting Information Figure 1).23 The octahedron struts are each composed of a bundle of six ~28 nm long double helices14,24 engineered with a ~90° curvature (see Supporting Information Note 1).15 DNOs were self-assembled in a one-pot reaction by combining phage-derived scaffold DNA (p7308)25 with 144...
oligonucleotide staple strands (see Supporting Information Table 1) in a 15 h thermal annealing ramp. Correctly folded structures were isolated by glycerol–

$\text{C}_0$

gradient centrifugation and verified by negative-

stain transmission electron microscopy (TEM) (Figure 1b and Supporting Information Figure 2).

Functional molecular features coupled to oligonucleotides can be assembled onto DNA nanostructures with high precision through hybridization to single-stranded DNA “handles” designed into the nanostructure. We designed an “inner” set of 12 identical handles (protruding from the inside face of the DNA struts) for attachment of fluorophore-conjugated oligonucleotides, providing optical contrast agent functionality. We designed a second “outer” set of 48 handles (protruding from the outside face) for attachment of lipid-conjugated oligonucleotides$^{27}$ with the lipid positioned $\sim 5$ nm (2 nm 6-thymidine spacer + 2.5 nm double helix width + 0.5 nm linker for lipid) from the nanostructure frame to drive tight wrapping of the membrane around the structure.

Our encapsulation strategy involves directing lipid bilayer assembly around DNA, recruited by individual lipid-conjugated oligonucleotides preassembled onto outer handles. The lipid bilayers are reconstituted out of a solution of mixed lipid and surfactant...
(N-octyl-β-D-glucopyranoside) through a dialysis step that selectively removes the surfactant to achieve the desired encapsulated DNO (E-DNO) (Figure 1c). We refer to DNO that has not been treated with lipids as non-encapsulated DNO (N-DNO). Unless otherwise stated, our membrane formulation includes 1,2-dioleoyl-sn-glycero-3-phosphocholine (DOPC) (94.2% molar contribution), 1,2-dioleoyl-sn-glycero-3-phosphoethanolamine-N-[methoxy(polyethylene glycol)-2000] (PEG-DOPE) (5%), and fluorescent rhodamine-1,2-dioleoyl-sn-glycero-3-phosphoethanolamine (Rh-DOPC) (0.8%).

We measured internalized DNO diameters of 53 nm versus outer membrane diameters of 76 ± 4 nm from TEM images of E-DNO (Supporting Information Figure 3). Our membrane formulation has an estimated thickness of ∼4 nm lipid bilayer plus ∼5 nm of PEG on each leaflet; therefore, the observed differential of ∼12 nm per vesicle side is consistent with a unilamellar envelope. As anticipated based on our design inspiration, the ultrathin dimensions of E-DNO are similar to enveloped virus particles (e.g., phi12 bacteriophage, Figure 1d) with our nanostructure taking the place of the protein capsid shell, wrapped tightly by a single lipid bilayer.

To optimize our bilayer reconstitution strategy, we synthesized additional variants of DNO having 0, 12, or 24 outer handles. After subjecting these to the lipid treatment described above, we examined the extent of association between nanostructures and vesicles using a membrane-impermeable DNA dye (PicoGreen). Fluorescence-based quantification of DNO before and after membrane destabilization with surfactant revealed a direct correlation between outer handle number and the percent of nanostructures inaccessible to the dye (Figure 2a). We observed 20.2 ± 3.5% (±SEM) of the 0 handle variant was membrane-enclosed, suggesting there is a nonspecific interaction between the DNA and lipids. This could be tuned by addition of negatively charged lipid (15% 1,2-dioleoyl-sn-glycero-3-phospho-L-serine) to the membrane formulation, which decreased the nonspecific interaction to 5.1 ± 2.7% (Supporting Information Figure 4). The variant with 12 outer handles at an average density of ∼1 handle per 710 nm² of nanostructure surface (surface area is defined as that of a sphere enclosing the DNO) showed 43.9 ± 3.1% membrane-enclosed. A comparable 48.3 ± 3.5% was measured with 24 handles (1 handle per ∼350 nm²). The association between DNO and membranes was strongest with our design maximum of 48 outer handles (1 handle per 180 nm²/handle), which showed 68.1 ± 7.3% enclosed within membranes.

We then tested protection from nuclease activity via DNase I digestion (Figure 2a). Following lipid treatment, 1.5 μg of each variant was incubated with 24 units of enzyme for 24 h at 37 °C, and total remaining DNA was quantified by PicoGreen fluorescence and compared to non-nuclease-treated controls. The 0 outer handle sample showed little protection with 30.4 ± 2.3% DNA remaining, whereas 84.6 ± 7.2% remained in the 48 outer handle sample, clearly...
showing that the envelope provides protection against nuclease digestion.

Negative-stain TEM revealed a far more dramatic impact of handle number on ultrastructure (Supporting Information Figure 5). Whereas most 0 handle DNOs were not associated with vesicles, those with 12 or 24 outer handles had much of their surface covered by vesicles formed external to the nanostructure frame. Imaging of 48 handle DNO revealed nanostructures within apparently complete and tightly associated lipid bilayers, as shown in Figure 1c. The number and density of attached lipid-conjugated oligonucleotides is therefore a key design parameter for DNO encapsulation. Successful encapsulation was also dependent on the liposome formulation. Removal of PEG-DOPA resulted in much larger reconstituted vesicles and no tightly wrapped nanostructures, suggesting that PEG plays an important role in controlling the extent of micelle fusion. We observed that cholesterol (∼5%) can be used in place of PEG-DOPA to similarly control micelle fusion and to produce E-DNO (data not shown). Inclusion of both PEG (5%) and cholesterol (15%) caused stabilization of reconstituted vesicles at smaller diameters, suggesting an inhibition of fusion and trapping of a putative intermediate of the encapsulation process (Supporting Information Figure 6).

We next used density equilibrium centrifugation in an iodixanol gradient to separate 48 outer handle E-DNO from excess lipid vesicles, resulting in a highly enriched population (Supporting Information Figure 7). The final recovery yield of E-DNO relative to starting number of DNO used in encapsulation and the fraction of fully encapsulated nanostructures were determined by Pico-Green quantification and membrane exclusion to be ∼20 and 90%, respectively. This yield allowed us to prepare quantities of E-DNO suitable for in vitro and in vivo studies.

A significant innate inflammatory response would be a serious impediment to many biomedical applications of DNA nanotechnology. As a gauge of this, we incubated Cy-5-labeled N- and E-DNO, as well as empty 50 nm vesicles, with immune cells isolated from mouse spleens. Similar to previously reported results,17,18 N-DNO activated a potent inflammatory cytokine response comparable to that produced from exposure to bacterial or viral nucleic acids. Interleuken-6 (IL-6) and IL-12 were produced at 136 ± 10- and 99 ± 5-fold above nonactivated cells (Figure 2b,c). In contrast, E-DNO and 50 nm vesicles induced 2.1 ± 0.3- and 0.9 ± 0.3-fold increases, respectively, in IL-6 production versus controls. Similarly, IL-12 was undetectable after incubation with E-DNO and vesicles. Flow cytometry showed that mean fluorescence of splenocytes incubated with N-DNO was 111 ± 8-fold higher than with E-DNO, which was equivalent to negative control cells (Figure 2d). N-DNO uptake was concentrated in a subpopulation of large, granular cells (Figure 2e,f) in which 89.0 ± 3.2% displayed very bright Cy5 signal. This same subpopulation showed low mean fluorescence and few positive cells (8.0 ± 2.3%) after incubation with E-DNO, a result that was confirmed by confocal microscopy (Supporting Information Figure 8). This splenocyte assay shows that nanostructure activation of, and uptake by, immune cells can be almost fully attenuated by encapsulation in a PEGylated lipid membrane.

Encouraged by this, we next aimed to characterize the in vivo pharmacokinetics of E-DNO. AlexaFluor750-labeled oligonucleotides, N-DNO, and E-DNO were intravenously injected into anaesthetized mice to allow tracking via whole animal optical imaging for 120 min postinjection. Elimination half-life of the oligonucleotide control was estimated to be 38.0 ± 0.8 min, with signal rapidly accumulating in the bladder immediately after injection (Figure 3a,b). The half-life (49.5 ± 1.0 min) and clearance pattern of N-DNO were equivalent to that of the oligonucleotide. Because nanoparticles larger than 6 nm are size-excluded from renal clearance,30 these data indicate that the structural integrity of N-DNO becomes compromised immediately after injection. In contrast, E-DNO displayed an estimated half-life of 370 ± 38 min (Figure 3c), comparable to similarly formulated PEGylated liposomes,4 with little bladder accumulation. In comparison to N-DNO, encapsulation increased the elimination half-life and relative bioavailability by factors of 9 and 17, respectively.

The stability of DNA nanostructures and devices has previously been probed during exposure to lysate from sodium dodecyl sulfate-treated mammalian cells,31 and after direct injection into Caenorhabditis elegans,32 with little degradation observed. Comparison to our findings is confounded by the presence of nuclease-inhibiting surfactant, distinct nanostructure designs, and an alternative in vivo model system, yet the observed differences suggest that DNA-based nanomaterials may be sensitive to their environment in a design-dependent manner. A folate-targeted, DNA-based nanoparticle was also recently tested in a murine tumor model,33 displaying an elimination half-life (24.2 min) and kidney uptake similar to our non-encapsulated DNO, suggesting it may have suffered from a comparable, rapid degradation profile after injection.

Finally, we profiled biodistribution by imaging organs harvested 120 min postinjection (Figure 3d,e). No significant differences were observed between the distribution profiles of oligonucleotide and N-DNO on an organ-by-organ basis, and both displayed significant renal excretion with signal in urine of 86.3 ± 3.6 and 88.5 ± 2.9% (% of total, photons/g), respectively. In contrast, E-DNO urine accumulation reached only 11.0 ± 4.4% of AlexaFluor750 signal at this time point, while maintaining 85.0 ± 4.4% in blood. Dual labeling of the E-DNO membrane (rhodamine) and DNA nanostructure (AlexaFluor750) allowed us to compare biodistribution of both components through multiplex imaging (Supporting Information Figure 9).
No significant differences were observed between the two components, suggesting that they remained associated throughout imaging.

As a last comparison, we examined the biodistribution of 50 nm liposomes of the same formulation, hydrodynamic diameter, and concentration as E-DNO (Extended Data Figure 9). No significant differences in rhodamine fluorescence were observed between the major organs. Yet surprisingly, the 50 nm liposomes cleared more rapidly than E-DNO, with twice as much signal (38.0 (3.2% vs 20.1 (4.3%, respectively) measured in urine. This was contrasted with 58.0 (2.5% and 76.2 (3.3% signal remaining in blood of liposome and E-DNO injected animals. Although it requires more in-depth follow-up studies, it would be interesting if DNO acts as a stabilizing endoskeleton and reduces clearance of liposomes.

CONCLUSIONS

This virus-inspired E-DNO displays favorable in vitro and in vivo properties, in stark contrast to non-enveloped DNO which activates a potent immune response and displays rapid degradation after injection. Using this design strategy as a starting point, many different biomedical applications can be conceived through integration of additional functions. Because our E-DNO incorporates a contrast agent and has appropriate dimensions and pharmacokinetic properties, it could easily be adapted for tumor detection. Addition of ligands to the outer membrane leaflet would add specificity for target molecules or cells, and such ligands could potentially be spatially organized by the DNA “capsid shell” using transmembrane linkers. Many different triggers could be implemented to drive an internal molecular cascade or a mechanical reconfiguration of the nanostructure frame (e.g., photon-fuelled,34 thermally35 or chemically induced36 conformational switch, pH-sensitive lipids37), which could be used for contrast agent unmasking, drug release, or more sophisticated behaviors that have been difficult to achieve using conventional nanomaterials or multicomponent nanodevices.9–11

A more distant but highly exciting prospect is the development of autonomous devices having functions approaching that of virus particles (e.g., cell entry via attachment and fusion) or primitive immune cells (e.g., input-based therapeutic response), and which could provide significant advances in diagnostics and therapeutics.

METHODS

Reagents. Lipids DOPC and DOPS, and PEG-PE, cholesterol and rhodamine-PE, the mini-extruder, extrusion membranes, and accessories were purchased from Avanti Polar Lipids. Endotoxin test cartridges (0.05–5.0 EU/mL) were purchased from Charles River. Accugene 10× TBE buffer, PCR tubes, and 96-well PCR plates (Axygen) were purchased from VWR. SYBR Safe and PicoGreen stains were purchased from Life Technologies.

Figure 3. In vivo optical imaging for analysis of pharmacokinetics and biodistribution. Mice were injected with AlexaFluor750-labeled oligonucleotide (orange), N-DNO (blue), or E-DNO dual-labeled with rh-DOPE (green) and imaged for 120 min postinjection. (a) Fluorescence images of mice at 120 min postinjection. E-DNO is seen throughout the body, whereas the other two agents have accumulated in the bladder (calibration bar = 500–20000 afu). (b) Mean fluorescence (measured in a region-of-interest traced around the head and torso) vs time, relative to the signal at t = 8 min (I/Io). (c) Elimination half-lives estimated from the kinetic analysis. (d) Fluorescence images of organs harvested 120 min postinjection, shown overlaid with photographic images (i) blood, (ii) urine, (iii) lung, (iv) heart, (v) liver, (vi) kidneys, (vii) spleen, calibration bar = 1500–25000 afu). (e) Organ distribution of AlexaFluor750 fluorescence (% of total, afu/g after correcting for calculated blood volumes of organs). (*a,b p < 0.05, ANOVA + Tukey’s test, dashed lines and error bars indicate SEM).
DNA Nanostructure Folding. The DNA nano-octahedron was designed using caDNAno.\textsuperscript{25} Stripes were produced reverse-phase-purified from Bioneer or Life Technologies Corporation. Fluor-coupled oligonucleotides were custom-synthesized and purified by IDT Technologies. The p7308 scaffold strand was produced from m13 phage replication in Escherichia coli, as described previously.\textsuperscript{26}

The scaffold strand was endotoxin-purified using Triton X-114. In brief, surfactant was added to scaffold stock to a final concentration of 2% (w/v) and incubated at 4 °C on an inversion mixer for 30 min. The solution was mixed at 37 °C for 5 min to cause phase separation, then centrifuged at 37 °C for 30 min at maximum speed in a benchtop centrifuge. The upper aqueous fraction was transferred to a new tube. This was repeated four times to reduce endotoxin in the scaffold solution to acceptable levels of less than 5 EU/mL, quantified using the Endotoxin-PTS system and test cartridges (Charles River). Downstream DNA nanostructure purification and dilution steps reduced endotoxin to undetectable levels prior to in vitro tissue culture and in vivo imaging experiments.

Folding was tested over a range of MgCl\textsubscript{2} concentrations. A solution of 10 mM scaffold was mixed with a 5-fold molar excess of each of 144 staples strands in 5 mM Tris (pH 8.0), 1 mM EDTA (1× TE), and 10–22 mM MgCl\textsubscript{2}. The solution was then subjected to a thermal annealing ramp on a Tetrad 2 Peltier thermal cycler (Bio-Rad) according to the following schedule: 80 °C for 5 min decrease to 65 °C at 5 min/°C; incubate at 65 °C for 20 min; decrease to 25 °C at 20 °C/min.

The products were separated in a 1.5% agarose gel with 10 mM MgCl\textsubscript{2} and SYBR Safe stain, in 0.5× TBE buffer + 10 mM MgCl\textsubscript{2} for 3 h at 60 V. The leading bands were extracted and imaged by TEM. We opted for a 14 mM MgCl\textsubscript{2} concentration for the synthesis of stock nanostructures.

Negative-Phase Transmission Electron Microscopy. TEM imaging was carried out by dropping 3.5 μL of product onto a plasma-treated carbon Formvar grid. This was incubated for 1 min, wicked away onto filter paper, 3.5 μL of 2% uranyl formate (in H\textsubscript{2}O, w/v) was added for 30 s, then wicked away. Imaging was carried out on a JEOl 1400 transmission electron microscope.

Purification of Synthesized Nanostructures. Solutions of folded DNA nanostructures were concentrated and then purified by glycerol gradient ultracentrifugation, as described elsewhere.\textsuperscript{26}

Liposome Preparation. Liposomes were prepared using standard techniques. We produced a mixture of DOPC, PEG2K-PE, and rhodamine-PE (molar ratio 94.2, 5.0, 0.8), or DOPC, DOPS, and octyl-β-D-glucopyranoside (OG), Tween20, magnesium chloride, magnesium sulfate, sodium chloride, and paraformaldehyde were purchased from Sigma-Aldrich, RPMI, PBS, FBS, and penicillin-streptomycin were purchased from Gibco. Carbon Formvar grids and uranyl formate were purchased from Electron Microscopy Sciences. The 96- and 384-well fluorescence assay plates were purchased from BD Biosciences. Amicon Ultra filtration devices, Optiprep media, and Seton ultracentrifugation tubes were purchased from Fisher Scientific.

DNA Nanostructure Encapsulation. In brief, DNA nanostructures were encapsulated by first annealing lipid–oligonucleotide and fluor–oligonucleotide conjugates to the nanostructure in a surfactant buffer. The annealed product was purified by glycerol gradient, then mixed with liposomes. This was then dialyzed to remove surfactant, and the product was purified, concentrated, dialyzed against a buffer appropriate to downstream experiments, and characterized. Sample volumes listed below were found to be scalable.

In a typical experiment, a 400 μL solution was prepared, containing 20–200 μg/mL DNO mixed with a 5× molar excess (relative to handle number) of lipid– and fluor–oligonucleotide conjugates in encapsulation buffer + 2% OG surfactant. The solution was incubated for a minimum of 2 h at 35 °C on a Tetrad 2 Peltier Cycler (Bio-Rad).

The annealed product was purified from excess lipid–DNA and fluor–DNA conjugates via glycerol gradients containing OG surfactant. Glycerol gradients were prepared using solutions of 15% glycerol + 2% OG and 45% glycerol + 2% OG in encapsulation buffer.\textsuperscript{27} The annealed product was layered on top of the gradients and centrifuged for 2.5 h at 41 000 rpm (for SW-41 rotor tubes). The gradients were then fractionated, and appropriate fractions were combined and concentrated back to the starting volume (i.e., 400 μL) using an Amicon 30K device.

The concentration of the DNA buffer was determined by UV absorbance at 260 nm on a Nanodrop system with disposable cuvettes. The solution was transferred into a 2.0 mL microcentrifuge tube. Liposomes were added by transferring a 0.5× volume of prepared liposomes into the solution (i.e., 200 μL). This was mixed on the Thermomixer at 450 rpm, room temperature for 1 h. A volume of encapsulation buffer equivalent to the current total volume (i.e., 600 μL) was added and mixed gently. The entire solution was then transferred into an appropriately sized 7K MTVCO Slide-a-Lyzer dialysis cassette (Thermo Scientific). The cassette was floated in 2 L of encapsulation buffer for 3 days for small samples (e.g., 120 μL) or 7 days for large volumes (e.g., 12 mL). Buffer was replaced every second day in all cases.

After dialysis, the sample was recovered from the dialysis cassette and concentrated using an Amicon column pretreated with encapsulation buffer.

Enveloped nanostructures were separated from excess lipids by equilibrium centrifugation using iodixanol (Optiprep reagent, Sigma-Aldrich). We prepared a working volume of 54% iodixanol from the stock 60% solution by mixing with 0.1× volume of 10× encapsulation buffer. We then mixed the 54% iodixanol/Encapsulation Buffer solution with 1× Encapsulation Buffer to prepare equal volumes of 35, 28, 18, and 8% iodixanol/encapsulation buffer. In the case of the 35% step, the encapsulated DNA nanostructure sample was used in place of 1× encapsulation buffer to dilute the 54% iodixanol solution. The volumes were layered into ultracentrifuge tubes and centrifuged at maximum allowable speed for 16 h, 4 °C. The gradient was fractionated, and 50 μL of each fraction was transferred into a 96-well fluorescence plate (BD Biosciences) and imaged on the Typhoon system (GE Healthcare Life Sciences). The fractionation was profiled by quantifying fluorescent signals in each fraction using ImageJ (NIH). The images were background-subtracted, then a circular region-of-interest (ROI) was placed over each well, and integrated density was measured. The integrated density values were plotted against fraction number to profile component distribution. Appropriate fractions containing E-DNO (based on Cy5 or AlexaFluor750 signal) were concentrated and washed with encapsulation buffer using an Amicon centrifugation filter device or, in some cases, were first dialyzed against a buffer appropriate for downstream experiments (e.g., phosphate-buffered saline, Gibco).

The encapsulated, purified products were characterized by TEM, DLS, and fluorescence, and the encapsulation yield was determined by a PicoGreen assay (described below). DLS was carried out on a Nano ZS (Malvern) using standard settings.
Fluorescence analysis was carried out on a Fluorolog (Horiba) using the following settings. For rhodamine fluorescence of vesicles, excitation was at 550 nm and emission scanned from 575 to 700 nm. For Cy5 nanostructure fluorescence, excitation was at 625 nm and emission scanned from 650 to 800 nm. For AlexaFluor750 fluorescence, excitation was at 700 nm and emission scanned at 725–900 nm.

**PicoGreen Dye Exclusion Assay.** The PicoGreen stain exclusion assay differentiates between membrane-enclosed and non-enclosed DNA based on the stain’s inability to cross a lipid bilayer. Measurements are made in parallel in encapsulation buffer and buffer with OG that destabilizes the membrane and allows the stain access to total DNA content of a sample. The assay is carried out in a 384-well black fluorescence assay plate (Greiner).

A standard curve was first prepared from the DNA nanostructure stock solutions. First 200 μL of 5 μg/mL was prepared in encapsulation buffer, then 6× 1:2 dilutions were prepared from this. Next 100 μL of each unknown sample was prepared by dilution with encapsulation buffer (typically 1.5–1.50×), aiming for the concentration of the diluted sample to be within the standard curve range. Two solutions of PicoGreen stain were prepared by diluting the stock reagent 1:200 in either encapsulation buffer, or encapsulation buffer + 2% OG. For each standard or sample to be measured, 10 μL of both PicoGreen solutions was pipetted into 3× wells of the 384-well plate. Ten microliters of the standards and samples was then added to the 6× wells, and the plate was incubated 5 min and light-protected. Fluorescence of the PicoGreen stain was measured on a SpectraMax M5 plate reader (Molecular Devices) by excitation/emission at 480/520 nm.

For analysis, standard curves were plotted for median fluorescence values of both the OG-negative and OG-positive buffers. These were used to calculate the DNA concentration of unknown samples in the two buffers. The concentration of DNA in the OG-containing buffer equals “total DNA”, whereas in the encapsulation buffer was “non-encapsulated DNA”. Encapsulation yield is then calculated by

\[
\text{Encapsulation yield} = \frac{\text{total DNA} - \text{non-encapsulated DNA}}{\text{total DNA} \times 100%}
\]

**Nuclease Protection Assay.** Sensitivity to nuclease activity was determined using DNase I (New England Biolabs). The encapsulation process was carried out on 0, 12, 24, and 48 outer handle DNA nanostructure variants at a point prior to purification by float-up centrifugation. One hundred microliters of each variant, 12 tubes in total). Ten units of DNase I were added to 3 μL of each variant, and an equivalent volume of DNase I buffer was added to 3× replicates as negative controls. All samples were incubated for 24 h at 37 °C on a Tetrac 2 Peltier thermocycler. Following this, the PicoGreen assay was used to determine the DNA remaining after nuclease digestion. The percent remaining was expressed as a ratio of the calculated DNA remaining in the DNase I-positive samples over the DNase I-negative samples for each design.

**Splenocyte Activation Assay.** RPMI media (Gibco) used for splenocyte culture were adjusted to maintain the stability of DNA nanostructures. MgSO4 was adjusted to 6 mM by addition of MgCl2 solution to 1× to DNase I-negative of each variant, 12 tubes in total). Ten units of DNase I were added to 3× replicates to achieve each variant, and an equivalent volume of DNase I buffer was added to 3× replicates as negative controls. All samples were incubated for 24 h at 37 °C on a Tetrac 2 Peltier thermocycler. Following this, the PicoGreen assay was used to determine the DNA remaining after nuclease digestion. The percent remaining was expressed as a ratio of the calculated DNA remaining in the DNase I-positive samples over the DNase I-negative samples for each design.

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tors of the animal to just below the front legs. The integrated density was measured in each image. The data set was normalized to the value at 8 min postinjection, as this showed peak fluorescence after diffusion of the agents throughout the vascular system. Organ measurements were obtained using the Freehand Selection tool to draw ROI around their perimeter and measuring integrated density. Fifty microliter volumes of blood and urine were measured. “Organ distribution” was calculated for each organ as a percent of the total measured arbitrary fluorescent units, with urine normalized to a 200 μL estimated bladder volume and blood to 2500 μL estimated total blood volume. The contribution of organ fluorescence from blood volume was calculated for each organ as a percent of the total measured fluorescence.

Statistics. Student’s t test and ANOVA with posthoc Dunnett’s or Tukey’s tests were performed using an excel plug-in, iner-
STAT-A v1.3 by Mario H. Vargas (Instituto Nacional de Enfermedades Respiratorias, Mexico).

Animal Use. All animal studies were performed in accordance with NIH guidelines, under approval of Harvard University’s Institutional Animal Care and Use Committee.

Conflict of Interest. The authors declare the following competing financial interest(s): We filed a provisional patent on this technology.

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Supporting Information Available: Methods, additional data, and a nanostructure design schematic and oligonucleotide sequences are available as Supporting Information. This material is available free of charge via the Internet at http://pubs.acs.org.

REFERENCES AND NOTES


