Mechanisms of Tissue Uptake and Retention in Zotarolimus-Coated Balloon Therapy

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Background—Drug-coated balloons are increasingly used for peripheral vascular disease, and, yet, mechanisms of tissue uptake and retention remain poorly characterized. Most systems to date have used paclitaxel, touting its propensity to associate with various excipients that can optimize its transfer and retention. We examined zotarolimus pharmacokinetics.

Methods and Results—Animal studies, bench-top experiments, and computational modeling were integrated to quantify arterial distribution after zotarolimus-coated balloon use. Drug diffusivity and binding parameters for use in computational modeling were estimated from the kinetics of zotarolimus uptake into excised porcine femoral artery specimens immersed in radiolabeled drug solutions. Like paclitaxel, zotarolimus exhibited high partitioning into the arterial wall. Exposure of intimal tissue to drug revealed differential distribution patterns, with zotarolimus concentration decreasing with transmural depth as opposed to the multiple peaks displayed by paclitaxel. Drug release kinetics was measured by inflating zotarolimus-coated balloons in whole blood. In vivo drug uptake in swine arteries increased with inflation time but not with balloon size. Simulations coupling transmural diffusion and reversible binding to tissue proteins predicted arterial distribution that correlated with in vivo uptake. Diffusion governed drug distribution soon after balloon expansion, but binding determined drug retention.

Conclusions—A large bolus of zotarolimus releases during balloon inflation, some of which pervades the tissue, and a fraction of the remaining drug adheres to the tissue–lumen interface. As a result, the duration of delivery modulates tissue uptake where diffusion and reversible binding to tissue proteins determine drug transport and retention, respectively. (Circulation. 2013;127:2047-2055.)

Key Words: animal model ■ binding ■ computational modeling ■ diffusion ■ drug-coated balloon ■ peripheral vascular disease ■ zotarolimus

Peripheral artery disease remains a clinical challenge despite advances in angioplasty and stenting.1,2 Intervention with the use of drug-coated balloons (DCBs) is emerging as a potentially viable strategy,3 demonstrating clinical efficacy at inhibiting restenosis after angioplasty in the lower extremities.3,4 The use of DCBs can open occluded vessels and concomitantly deliver drug to target lesions while avoiding the risks of chronic inflammation and incomplete healing associated with permanent implants such as stents. Most studies to date using DCBs have used paclitaxel, with a range of hydrophilic carriers and coatings, and have demonstrated varying levels of efficacy.4–7 Recently, zotarolimus-coated balloons (ZCBs) have shown efficacy within hypercholesterolemic femoral arteries in swine.5 Although positive results have been observed in peripheral applications, the use of DCBs in de novo or ST-segment elevation myocardial infarction coronary lesions in conjunction with a bare metal stent has shown a lack of efficacy versus a conventional drug-eluting stent.9,10 Safety concerns raised by these data illustrate a need for better understanding of drug distribution and residence time. Debate continues as to whether DCBs induce sustained, desirable clinical effects and whether bolus release of drug at the lumen–tissue (or mural) interface during balloon expansion is a clinically efficacious mode of delivery.

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We sought to determine the mechanisms that govern arterial uptake and distribution of balloon-delivered drugs with a focus on zotarolimus. By the use of an integrative framework coupling in vivo, bench-top, and computational models
of ZCBs deployed in porcine arteries, we quantified spatiotemporal arterial drug distribution patterns. Tissue incubation experiments using radiolabeled zotarolimus and paclitaxel provided estimates of the partition constants, tissue-binding capacities, and transmural drug diffusivities, with paclitaxel serving as a benchmark during experimental determination of the transport constants. In vitro release experiments of ZCBs expanded for different durations in porcine blood estimated the amount of releasable drug that is transferred to the tissue during balloon expansion. In vivo studies using ZCBs exposed for 2 inflation times revealed a dependence of tissue content on delivery duration. Simulations performed on a computational model constructed by using the bench-top parameter estimates revealed that the time-varying arterial drug distribution patterns resulting from balloon delivery are governed by a delicate balance between diffusion-mediated drug transport into the arterial wall and reversible binding to tissue ultrastructural elements.

Materials and Methods

Net Partition Constant and Binding Parameters

Arterial partition constants of zotarolimus and paclitaxel dissolved in phosphate-buffered saline were measured. Unlabeled and radiolabeled analogs of each drug were mixed at a ratio of 100:1, because this provided sufficient signal for measuring tissue concentrations via liquid scintillation counting. Drug bath solutions were prepared at 5, 10, and 20 µmol/L concentrations in phosphate-buffered saline (pH 7.4) containing 4% (wt/vol) bovine serum albumin and stored in glass vials at 4°C. Two percent polyethylene glycol 200 (wt/wt) was added to all solutions to improve the aqueous solubility of the drugs and prevent nonspecific adhesion to glass.11 Unlabeled and radiolabeled zotarolimus ([3H]Zot) were donated by Abbott Vascular (Santa Clara, CA) and Abbott Laboratories (Abbott Park, IL), respectively. Unlabeled paclitaxel was purchased from LC Laboratories (Woburn, MA), and radiolabeled paclitaxel ([3H]Pxl) from ViTrax (Placentia, CA). Porcine femoral arteries, immersed immediately in a 4°C bath of phosphate-buffered saline (pH 7.4) containing 4% bovine serum albumin, were obtained from a local slaughterhouse within hours of euthanization (Research 87, Boylston, MA). Arteries were cleaned of excess fascia, cut into small cross-sectional segments (20–60 mg), and placed in glass vials with drug bath solutions at 4°C for different incubation times up to 96 hours. All samples were placed on a shaker at 10 to 20 rpm to homogenize the drug bath and facilitate uniform exposure to tissue. After incubation, each tissue sample was immersed in a separate glass vial containing 1 mL of an aqueous solubilizer (Solvable, PerkinElmer Inc.) and dissolved at 55°C with shaking for 24 hours. Two hundred microliters of the dissolved liquid from each vial processed in triplicate was combined with a pseudocumene-based cocktail (Hionic-Fluor, PerkinElmer Inc) for liquid scintillation counting. The net partition constant (κ) was then defined as the ratio of the relative vial concentration of the digested tissue sample appropriately normalized with tissue sample weight ($C_v$ in micromoles per liter) and the drug bath concentration ($C_{bulk}$ in micromoles per liter).

Net tissue-binding capacity ($B_t$ in micromoles per liter) and equilibrium dissociation constant ($K_d$ in micromoles per liter) were estimated by varying the concentration of drug solutions (5–20 µmol/L) and considering the variations in κ for 96 hours. Experimental partition constant values were then fit (GraphPad Prism 3.0) to the relationship implied by bimolecular binding of small hydrophobic drugs that have access to the entire tissue volume:12

$$\kappa = 1 + \frac{B_t}{K_d + C_{bulk}} \quad (1)$$

Transmural Distribution and Effective Diffusion Coefficient

Intra-arterial distributions of zotarolimus and paclitaxel were measured using a diffusion chamber (Harvard Apparatus) (Figure 1A). To minimize hydrophobic drug binding to the interior walls of the diffusion chamber, hydrophilic surface modification was performed by incubating the chamber in 10 µmol/L poly(lysine) for 6 hours at room temperature. Porcine femoral arteries prepared as described above were cut into small segments ($≈2×2$ cm²) and affixed between the opposing faces of the wells within the diffusion chamber by using a series of pins surrounding the opening. The half-chambers comprised baths containing 10 µmol/L drug dissolved in phosphate-buffered saline+$4\%$ bovine serum albumin+$2\%$ polyethylene glycol 200; drug-free baths were used as controls. Tissue samples were maintained at a steady bath concentration gradient for 1 hour at room temperature. Magnetic stir bars were used throughout the incubation to ensure adequate bath homogeneity. After incubation, samples were snap frozen with liquid nitrogen–cooled isopentane and immediately stored at −80°C until further analysis.

The mass of each frozen segment was recorded, and 20-µm sections were cut parallel to the intima with a cryotome (Leica Inc). Triplets of consecutive sections were placed together in a glass vial containing 1 mL of Solvable and subsequently digested at 55°C for 24 hours. Liquid scintillation counting was used to measure drug

![Figure 1](https://example.com/f1.png)
concentration in each vial containing dissolved tissue and Hionic-Fluor. Net tissue concentration in each vial was then normalized to the highest estimated mean concentration from all the vials and plotted by using the midpoint of each triplet of sections as the transmural location on the abscissa (Figure 2B). For each distribution profile, permeation depth \( (L_p) \), denoted as the distance at which the ratio of intra-arterial tissue and the source concentrations at the mural surface reaches 0.083 (see online-only Data Supplement) was calculated, and the effective diffusivity \( (D_{eff}) \) of the total drug within tissue was estimated from the equation:

\[
D_{eff} = \frac{I_p^2}{6t}
\]

where \( t \) is the total time of incubation; \( D_p \) represents apparent net diffusivity of free and bound drug within the arterial tissue, which is lower than diffusivity in fluid because of the combined effects of steric hindrance and reversible binding to immobilized proteins. As tissues were exposed to low drug concentrations, the diffusivity of free drug \( (D_p) \) was estimated by using the result for unsaturated drug binding (see online-only Data Supplement):

**Transfer Kinetics**

The amount of releasable zotarolimus during balloon expansion was measured with ZCBs (6×40 mm; dose, 300 mg/cm²; Figure 1B), in a bench-top system. A 250-mL beaker filled with citrated porcine blood, gently stirred, and maintained at 37°C was used to maintain sink conditions. A fresh aliquot of blood was used for each balloon. A ZCB was placed into the solution, inflated to 2 atm, and held in solution for the designated time. The balloon was immediately removed, followed by release of pressure on the balloon and excision of the balloon from the catheter. Extracted balloons were then placed in individual vials and rapidly frozen at −20°C. Experiments were performed in triplicate at 4 different expansion times (30–300 seconds) along with a control arm of zero expansion time. For all balloons, drug content was quantified via high-performance liquid chromatography after extraction (column, Zorbax Eclipse XDB-C18, 3.5 μm, 4.6×50 mm; temperature, 60°C; flow rate, 1.2 mL/min; mobile phase, 10 mmol/L ammonium acetate buffer/acetonitrile gradient; injection volume, 10 μL; detection, A(w)). The drug released was calculated as the difference relative to control ZCBs. The amount of zotarolimus released from balloons was normalized to the area of the balloon \( (M_b) \) which was fit to 1-phase exponential association kinetics as follows:

\[
M_b(t) = A_1(1 - e^{-kt})
\]

Here, \( t \) is the time and \( A_1 \) and \( k \) are empirically estimated by using curve fitting (GraphPad Prism 3.0). The flux \( (J_b) \) then becomes:

\[
J_b(t) = (k_1A_1/Z_{MW})e^{kt},
\]

where \( Z_{MW} = 966.21 \) g/mol is the molecular weight of Zotarolimus.

**In Vivo Tissue Uptake**

Tissue uptake and retention of zotarolimus was estimated in domestic farm swine of either sex (30±5 kg, n=18). All animals received standard care in accordance with the Animal Welfare Act and the “Principles of Care of Laboratory Animals” formulated by the Institute of Laboratory Animal Resources (National Research Council, National Institutes of Health publication No.85-23, revised 1996); Clupidogrel (150 mg) and aspirin (325 mg), and enrofloxacin (5 mg/kg) were administered 1 day before and on the day of the procedure. Nifedipine (30 mg) was provided 1 day before the procedure in the evening and on the day of the procedure in the morning. Buprenorphine (at 0.01–0.06 mg/kg) was administered as needed for pain management; 8F femoral or iliac arterial access was obtained. After heparinization, each animal was treated with 4 balloons in the external iliac and the superficial femoral arteries and profunda arteries (Figure I in the online-only Data Supplement). The balloon coating composed of zotarolimus combined with proprietary excipients at a weight fraction of ±0.59. The exact location of the treatment site of the artery was identified based on angiograms with the use of anatomic landmarks as references and a scaling factor calculated based on actual and measured balloon length.

Arteries were exposed to 30±1 second or 180±2 second inflations (Figure I in the online-only Data Supplement). Fox sv PTA catheters (Abbott Vascular, CA) with drug dose (300 μg/cm²) available in 2 sizes, 5×40 mm (1.88 mg/balloon) and 6×40 mm (2.26 mg/balloon), were used to enable appropriate sizing and target balloon-artery expansion ratio (1.2:1). Overall mean swine arterial diameter in the study was 4.8±0.6 mm. Two hundred micrograms of nitroglycerin was administered intraluminally as needed to control arterial vasospasm. Animals were euthanized after 5 minutes, 4 hours, and 24 hours; treated arteries were excised and collected based on angiography data with the use of anatomic landmarks as references. At the end of the respective time points, vessels from each animal were dissected out in the same order as that of the treatment, and the arteries were carefully cleaned from surrounding connective tissue. Cleaned arteries were placed in glass vials that were snap frozen in liquid nitrogen and stored at −80°C. For analysis of the zotarolimus
content, each arterial tissue specimen was thawed, weighed, and processed for assaying by liquid chromatography coupled with mass spectroscopic detection.14

Computational Model

Transient drug distribution from an expanded DCB was modeled as a 2-dimensional continuum transport problem. The computational domain constituted an arterial cross-section with radius (R=3 mm) and wall thickness (T=0.5 mm) (Figure 1C). Free drug was allowed to diffuse with a constant diffusivity (Df) and reversibly bind to nonspecific tissue sites according to the local balance equations12:

\[ \frac{\partial C_f}{\partial t} = D_f \nabla^2 C_f - \frac{\partial B}{\partial t} \]

and

\[ \frac{\partial B}{\partial t} = k_s C_f (B_M - B) - k_d B \]

The variables C_f and B denote the local concentrations of free and bound drug in the arterial wall, respectively, B_M is the net tissue-binding capacity, and k_s and k_d are the association and dissociation rate constants, respectively. These equations were solved subject to zero initial free and bound drug concentrations within the tissue, flux boundary condition during balloon inflation at the mural surface, and perfect sink condition at the adventitial surface. At the mural surface, drug influx is prescribed as:

\[ J_b(t) = \begin{cases} J_b(t), & t \leq t_0 \\ 0, & t > t_0 \end{cases} \]

where t_0 is the balloon inflation time and J_b(t) is the flux approximating the releasable portion of zotarolimus from the balloon during inflation (Equation 5). Once the balloon has been deflated and retracted, bulk transferred drug at the mural surface becomes exposed to flowing blood and can be cleared depending on its adherence to the wall and solubility in blood. Rather than modeling adherence and solubilization, we considered 2 opposing extremes: either that mural-adhered drug is insensitive to flowing blood, modeled as zero flux condition on this interface (Equation 8), or that blood is extremely efficient at clearing mural-adhered drug, modeled as a zero concentration condition:

\[ J_b(t) = J_b(t), \quad t \leq t_0 \quad \text{and} \quad C_f(t) = 0, \quad t > t_0. \]

Constants including B_M, K_d, and D_f were estimated from the bench-top experiments. k_s and k_d were then computed as k_s = \frac{D_f D_a}{B_M T} and k_d = k_s K_d. Here, D_a = 50 000 is the Damkohler number based on rapamycin data.12 Time-dependent simulations (COMSOL Inc) were performed on the computational domain that was meshed by using the Delaunay triangulation scheme. A zero concentration condition was applied on the perivascular side of the arterial wall for the free drug. For the bound drug, both the intramural and the perivascular aspects of the arterial wall were assigned a zero flux boundary condition. The direct (Sparse Object Oriented Linear Equations Solver) method was used to solve the system of equations with a nested dissection preordering algorithm, and a backward differentiation formula method was used for time integration with relative and absolute tolerances for the time stepping assigned at 10^{-4} and 10^{-8}, respectively. Simulations were performed until the Newton iterations of the fully coupled solver reached 1000 or when the minimum damping and tolerance factors reached 10^{-4} and 1, respectively. The methodology was deemed mesh independent when there was <2% difference in the mean free drug concentration within the arterial wall for successive mesh refinements, and the resultant mesh comprising 21 208 triangular elements was used for all subsequent simulations.

Statistical Analysis

Experimental data are expressed as mean±SEM. For transmural distribution data (Figure 2B), an F-test was performed to determine the goodness of fit of the linear distribution. When sample normality was justified, statistical comparisons between groups were performed with the unpaired student t test assuming unequal variances. When normality could not be supported, the 2-sample Mann–Whitney test was used. The probability values presented were derived from the t test unless otherwise indicated. Experimental differences were statistically significant at P<0.05.

Results

Arterial uptake of zotarolimus and paclitaxel greatly exceeded the applied bulk concentrations (Figure 2A), as reflected by their large partition constants (κ>1). Bath concentrations (5, 10, and 20 μmol/L) were selected to simulate high-concentration source conditions. Estimated net partition constants increased initially with time and approached steady state. For example, partition constant for zotarolimus increased >1.5-fold from 24 to 69 hours and remained relatively unchanged with further increase in time. Based on these observations, a bath concentration of 10 μmol/L and an incubation time of 69 hours were chosen for subsequent experiments to simulate saturating drug conditions. Curve fitting (Equation 1) on the distribution profiles of both drugs estimated κ under these conditions (9.3±0.4 for zotarolimus and 9.4±0.8 for paclitaxel) and provided estimates of the equilibrium binding parameters (B_M=0.356 mmol/mL and K_D=0.0326 mmol/L, R^2=0.999) for zotarolimus and (B_M=1.3 mmol/mL and K_D=0.136 mmol/L, R^2=0.515) for paclitaxel.

When drugs were exposed to the luminal side of the artery affixed within the diffusion chamber, zotarolimus uptake was maximal within the intimal region and decreased linearly with increasing distance from the intraluminal side (Figure 2B, linear regression and F-test: P<0.00005). In contrast, paclitaxel uptake displayed multiple peaks across the arterial wall (Figure 2B, linear regression and F-test: P>0.05). Although zotarolimus uptake was >13-fold higher at the intima than the uptake within the last transmurally grouped layer of the adventitia, paclitaxel uptake was only ≈3-fold higher. Estimated apparent net diffusivity (D_{app}) of paclitaxel (54.1±34.3 μm^2/s) was >3-fold higher than that of zotarolimus (17.1±5.1 μm^2/s). The true diffusivity (D_f) for unbound (free) zotarolimus within the tissue then becomes 204.1 μm^2/s (Equation 3).

Only ≈2% of total drug on the coating and ≈24% of the releasable drug is transferred during 30 seconds of inflation time (Figure 3A). Release followed first-order kinetics with a half-life of ≈75 seconds, where drug release from the balloon rapidly increased after balloon expansion and reached steady state within ≈5 minutes. This profile may reflect a solubilizing effect of a finite amount of excipient, which is subsequently depleted, or it may be attributed to the progressive smoothing of initially porous, high–surface-area morphology, slowing the release. In the presence of a monolithic, phase-separated coating, rapid release may reflect diffusion of a small amount of percolating drug on the surface of the balloon. Slower release afterward denotes the absence of sufficient time to allow the remaining drug to solubilize and diffuse through the coating.15 Micrographs of treated arteries illustrate that some zotarolimus released during balloon expansion adheres
directly to the mural surface and creates a thin layer of drug coating (Figure 3B).

Our data speak to bulk transfer of drug from the expanded balloon onto the mural surface followed by transport via diffusion and reversible binding within the arterial wall. Total drug uptake averaged over the treated tissue displayed biphasic kinetics, with an initial first-order decline phase during the first 4 hours that was succeeded by slower efflux of drug (Figure 4A and 4B). Approximately 98% of zotarolimus taken up by the artery was cleared between 5 minutes and 24 hours after 30-second balloon inflation. Yet, resulting arterial levels at 24 hours reflect detectable and potentially therapeutic levels of zotarolimus (1.4±0.5 ng/mg). Arteries exposed to 180-second balloon inflation exhibited higher tissue uptake than with 30-second exposure (Figure 4A and 4B) for superficial femoral and iliac arteries (Table I in the online-only Data Supplement), and tissue uptake was independent of balloon size (Table II in the online-only Data Supplement). Tissue uptake in profunda arteries at 4 hours was ≈7-fold (P<0.01) and ≈8-fold (P<0.01) higher than the superficial femoral and iliac arteries for 30- and 180-second balloon exposure times, respectively. Duration of balloon exposure to arteries impacted uptake in profunda arteries, as well, with 97% higher uptake observed at 4 hours (P<0.02), when inflation time rose 6-fold to 180 from 30 seconds.

Model simulations based on the bench-top–estimated binding parameters also predicted biphasic kinetics for total drug (Figure 4A and 4B), and the unbound and bound drug concentrations (Figure 5A). Predicted free-to-bound drug ratio becomes ≈9:10 and ≈1:1 after exposure to 30- and 180-second balloon inflations, respectively. This ratio increased exponentially in the first hour postexpansion and reached a steady state within the next few hours (Figure 5B). Simulations that varied only the drug diffusivity relative to in vitro estimates resulted in an envelope of distribution profiles with a trend similar to that of the in vivo measurements. For both inflation durations (Figure 4A and 4B), in vivo tissue uptake is consistent with the assumption that the thin layer of adhered drug that is left behind postballoon retraction shields subintimal drug from clearance by flowing blood (Equation 8). In contrast, simulations that assumed a highly permeable layer of mural-adhered drug underestimates the in vivo trend line (Figures II and III in the online-only Data Supplement).

Having demonstrated the correlation between computationally modeled and in vivo–derived tissue content, we...
Factors Governing Drug Uptake and Retention

Drug potency is not the sole determinant of the expected biological and ultimate clinical outcome after local delivery. Factors such as drug distribution and retention in and around target tissue become immensely important, and it is incumbent on us to understand their determinants. Our data illustrate the role of drug and tissue properties, duration of delivery, diffusion, and binding in determining zotarolimus uptake and retention in the underlying tissue.

Coating Composition

DCB coatings are typically drug rich and incorporate low-molecular-weight excipients aimed to facilitate drug release and transfer to the vessel wall. Vessel–balloon contact times are short, and the nature of contact during expansion and retraction, and the methods to modify coating composition, as well, to optimize balloon-transferred drug are still being defined. Thus, mechanisms governing drug transfer to the wall, distribution therein, and sustained retention have yet to be fully understood. This has led to competing mechanistic hypotheses as to the determinants of arterial distribution in DCB-treated arteries. Some argue that these processes are governed by transport and binding of soluble drug in the tissue, whereas others claim that a significant fraction of the transferred drug and excipients distribute in the artery in microparticulate form and that solubilization of these micro-depots determines tissue retention. From this perspective,
the nature of the coating determines the amount of drug that adheres to the mural surface and the degree to which subintimal soluble drug is protected by the mural-adhered coating from clearance by flowing blood. Our computational model validates a process that integrates these elements.

**Duration of Delivery**

Our in vivo results indicate that increasing arterial exposure time to an inflated balloon increased tissue uptake of zotarolimus. These observations are consistent with a first-order dependence of drug uptake on the duration of balloon inflation (Figure 3A) and the findings validated by computational simulations of zotarolimus arterial uptake (Figure 4A and 4B). Although prolonged balloon inflation might induce ischemia and arterial injury, the correlation between in vivo uptake data and the inflation time suggests the possibility of an optimal exposure time for ZCB-treated arteries. Penetration depth increased with balloon exposure time and drug was predominantly unbound at the end of respective inflation times. Shortly after balloon inflation, zotarolimus permeated efficiently throughout the arterial wall irrespective of exposure time. These results indicate that longer balloon exposure led to increased tissue uptake where diffusion and reversible binding to tissue proteins explain observed in vivo retention kinetics in ZCB-treated arteries.

**Drug Dependence**

When arteries were exposed to a constant source of drug in a diffusion chamber, transmural distribution of zotarolimus was maximal in the intima and consistently declined at increasing depths (Figure 2B). Such distribution is consistent with diffusion-mediated transport and a relatively uniform distribution of drug-binding sites across the artery wall. In contrast, under similar exposure, paclitaxel distribution displayed multiple peaks within the intimal and medial regions (Figure 2B).15,19 Indeed, the multilaminate structure of the arterial wall comprising elastin within the intimal and medial regions exhibits preferential binding to paclitaxel, and this phenomenon may explain, in part, its nonlinear distribution.20 Thus, although both these potent antiproliferative agents are small and hydrophobic, our data speak to their differential interactions with tissue proteins and are consistent with our previous finding for paclitaxel and sirolimus in calf carotid arteries.19,21 Because binding determines drug retention, our results suggest that paclitaxel and zotarolimus may also be differentially retained in arterial tissue. Sensitivity analysis showed that increasing effective diffusivity does not necessarily imply higher tissue uptake (Figure 4A and 4B; Figures II and III in the online-only Data Supplement). Higher diffusivity leads to faster transport and clearance that, in turn, places a threshold on tissue uptake by limiting drug availability for binding. Thus, parameters defining an efficient DCB delivery are bounded by a need for delivering a large amount of drug within a short period and the nature by which binding determines tissue retention.

**Early Dominance of Diffusion and the Role of Binding**

After balloon transfer of drug to the mural surface, zotarolimus permeates the entire tissue within a short period (Figure 6C) via diffusion. High values of diffusivity and the resulting penetration depth of zotarolimus not only ensure adequate and rapid arterial uptake during balloon expansion (Figure 6A and 6B), but also mediate faster drug clearance from the adventitial surface after balloon deflation. Consequently, greater arterial uptake comes at a cost of faster diffusive clearance, suggesting that drug diffusivity may be amenable to optimization. This might be achieved through the manipulation of the size or charge of the drug22 or through the choice of excipients or carriers that are added to the coating. This dual role of diffusion points to its dominating effect soon after balloon expansion as the drug within the arterial wall reaches a balanced state with perhaps seemingly adequate levels of concentration.

Binding plays a critical role in DCB therapy by protecting the drug from pervading out of the arterial wall. Our model results indicate that binding occurs immediately after drug exposure to tissue, and zotarolimus is mostly bound within 5 minutes of balloon expansion (Figure 6C and 6D). The ratio
of bound to unbound zotarolimus over time rises and reaches a steady-state value within 4 hours (Figure 4C), suggesting that binding determines long-term tissue concentration and retention. Consequently, strategies for increasing drug affinity to tissue proteins have the potential to significantly increase retention because factors including the drug’s binding potential and diffusivity modulate the balance between free and bound drug within the tissue.

**Tissue Dependence**

The roles of binding and diffusion introduce a dependence on tissue that can limit the extrapolation of clinical and preclinical experiences between vascular beds. Each arterial bed differs in form and element because muscular and elastic arteries are different from each other. Individual tissue layers within these vascular beds may allow isotropic diffusion; however, transport becomes anisotropic between alternating tissue layers of varying permeability. Certain arteries are anatomically structured to divide into several side branches, and, as a result, the area of the mural surface exposed to the inflated balloon is reduced. Such differences affect transport and our results point to a similar phenomenon. Specifically, profunda arteries retained significantly more drug at 4 hours than superficial femoral and iliac arteries. This difference might reflect differential permeability or binding capacity owing to ultrastructural differences, or differences in adventitial washout owing to the embedding of profunda arteries within dense muscle tissue. It may also reflect the inherent nonbifurcating nature of the profunda artery where a larger surface area is covered by the expanded balloon without drug loss into the side branches.

**Arterial Wall Composition**

Atherosclerotic vessels are composed of nonuniform layers of lipid, calcification, and other inflammatory agents that create a heterogeneous wall ultrastructure. Balloon expansion within these vessels likely has differential effects that reflect different morphologies and compositions of the target tissue. Kinetics of drug transfer within a localized region of injury may be altered, and these changes can lead to nonuniformities in drug distribution. Increasingly sophisticated disease animal models are being presented, and yet the variability among each arterial segment, vascular bed, and animal currently challenges the correlations we seek, especially as we focus on quantifying microlevel tissue drug distribution. Nonetheless, we have previously observed that net tissue uptake into atheromatous rabbit arteries can mask significant differences in the tissue distribution patterns of paclitaxel and sirolimus derivatives. Thus, the overall correlation between the in vivo tissue content and the computational model–based predictions may very well carry into diseased animal and human arteries. Future studies should examine the correlations between models and in vivo data on a microscopic level where variations in tissue morphology are apparent in a given specimen and from 1 vessel to the other.

**Study Limitations and Future Directions**

An integrative approach coupling in vivo, computational, and bench-top models allowed us to understand the mechanisms governing zotarolimus transfer from coated endovascular balloons and subsequent arterial distribution and retention. A 2-dimensional computational model provided estimates of drug retention and binding at different time points. This model allowed empirical estimation of drug transport within tissue and simulation of the complex phenomenon of reversible binding within the arterial wall. However, assumptions were made to simplify our understanding and create a coherent connection between bench-top–derived pharmacological constants that were fed into the computational model and the model-based predictions that matched with the in vivo data. For example, 2 sets of boundary conditions approximated 2 opposing scenarios: the case in which flowing blood completely washes out the mural adhered drug and the case when the adhered drug is insensitive to flow. Future studies will be performed with the use of higher-fidelity models using physiologically realistic geometries, anatomically meaningful arterial wall composition, and the rate at which mural-adhered drug is cleared by systematically accounting the role of luminal flow.

Our bench-top, animal, and computational models do not account for the presence of atherosclerotic plaque and calcification. Compositional changes in the artery that accompany increased atherosclerosis affect local tissue capacity for drug absorption and retention. Because disease-induced changes alter the distribution of drug-binding proteins and interstitial lipid alters drug distribution, one might observe differences in healthy versus normal arteries and, hence, tissue saturation of drug might be comparative but not similar. Moreover, it has been shown that sirolimus derivatives are relatively less sensitive to lesion complexity than paclitaxel, suggesting that the mechanisms of ZCB therapy explained in this study may remain intact, as well, for diseased vessels.

**Conclusions**

DCBs can deliver to and sustain drug within the arterial wall but in a manner distinct from permanent implants such as drug-eluting stents. A large bolus of drug released from an endovascular balloon over a limited time adheres to the mural interface, some of which diffuses into the tissue and saturates the tissue-binding sites up to a well-defined penetration front. A delicate balance between diffusion, binding, and clearance at the mural and adventitial interfaces determines subsequent distribution and retention. This balance depends on the duration of delivery, the degree to which the adherent layer of mural drug shields subintimal soluble drug from blood washout, and the binding capacity of the arterial bed, and it is amenable to optimization via manipulation of drug diffusivity and nonspecific binding properties.

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References


CLINICAL PERSPECTIVE
Local drug delivery from endovascular balloons investigated decades ago has been rejuvenated with the expectation that issues like thrombosis with drug-eluting stents and late lumen loss with bare metal stents could be avoided. Early failures of heparin-eluting catheters and balloons were attributed to poor retention of hydrophilic drugs, and, indeed, hydrophobic paclitaxel is retained, because it associates with hydrophilic carriers to enhance the transfer from blood to the artery wall and retains when dissociated. It remained unclear, however, whether sirolimus derivatives such as zotarolimus that are efficacious when released from drug-eluting stent but may not use the same transport-enhancing mechanisms as paclitaxel, could demonstrate comparable efficacy when coated on balloons. Our work is the first to describe the mechanisms of zotarolimus-coated balloon therapy by the use of an integrative approach coupling in vivo studies, bench-top experiments, and computational modeling. A large bolus of balloon-released zotarolimus and its constituents transfer during inflation, some drug pervades the tissue, and a fraction of the drug coating adheres to the tissue–lumen interface. The duration of balloon exposure to the tissue–lumen interface determines the net drug uptake into tissue, where diffusion mediates transport into the arterial wall and reversible binding to tissue ultrastructural elements determines the retention of zotarolimus in an arterial bed–dependent manner. Therefore, there is a theoretical basis for balloon delivery of zotarolimus to the arterial wall to be clinically efficacious and that optimization of zotarolimus-coated balloon therapy may rely on the tailoring of balloon coating, drug release kinetics, and inflation time to the arterial target.
Mechanisms of Tissue Uptake and Retention in Zotarolimus-Coated Balloon Therapy

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Estimating the apparent and true diffusivities in arterial tissue

Tissue drug uptake and retention are determined by a combination of transport into the arterial wall and binding to tissue ultrastructural elements\textsuperscript{1-4}. For most drugs, including Sirolimus derivatives and Paclitaxel, binding within the tissue is diffusion limited and manifests as a concentration dependent effective diffusion coefficient:\textsuperscript{5}

\[
D_{\text{eff}} (C_f) = \frac{D_T}{1 + B_p/(1 + C_f/K_d)^2}. \tag{S1}
\]

Here $B_p$ is the binding potential of the tissue for the particular drug, $C_f$ is the local concentration of the free drug in the tissue, $D_T$ is the diffusion coefficient of the free drug in the tissue and $K_d$ is the equilibrium dissociation constant of drug binding to tissue elements. This relationship accounts for the fact that drug hindrance due to binding is minimal at saturating concentrations and maximal in the opposite extreme of binding site excess over drug dose. The latter limit is achieved for free drug concentrations that are low relative to the dissociation constant. Specifically, when $C_f < K_d/3$ the following approximation is valid to within a 10% error:\textsuperscript{5}

\[
D_{\text{eff}} (C_f) \approx \frac{D_T}{1 + B_p}. \tag{S2}
\]

To avoid the confounding effects of concentration on estimated diffusivities, we ensured that ambient drug concentrations in our diffusion chamber experiments were low relative to the respective dissociation constants. To further simplify the analysis and highlight the sensitivity to diffusion, we measured drug distribution after 1 hr as such short exposures result in partial penetration of the drug. Under such conditions, the following distribution of drug is predicted for one-sided exposure of a flat tissue segment with uniformly distributed drug binding sites:\textsuperscript{5,6}

\[
C_f (x) + B(x) \approx (1 + B_p) \cdot C_f (0) \cdot \text{erfc} \left( \frac{x}{2 \sqrt{D_T t/(1 + B_p)}} \right). \tag{S3}
\]
In analyzing experimental data, it is convenient to normalize the concentration relative to the surface drug concentration:

\[
\frac{C_t(x) + B(x)}{C_t(0) + B(0)} = \text{erfc}\left(\frac{x}{2\sqrt{D_t t/(1 + B_p)}}\right) \tag{[S4]}
\]

Defining the penetration depth \((L_p)\) as the distance at which tissue drug concentration drops to 8.3% of its surface value estimated the effective diffusion coefficient as:

\[
D_T/(1 + B_p) = \frac{L_p^2}{6t} \tag{[S5]}
\]

and the true diffusivity as

\[
D_T = (1 + B_p) \times \frac{L_p^2}{6t}. \tag{[S6]}
\]

### Tissue-specific drug interactions

Bench-top estimated parameters such as maximum tissue binding capacity \((B_m)\) and equilibrium dissociation constant \((K_d)\) speak to the ultrastructural tissue elemental concentrations of the binding sites and the molecular interactions of a specific drug, respectively. While these parameters provide quantitative perspectives on the tissue state and drug’s binding affinity, they lack the ability to quantitatively define tissue specific drug interactions. We therefore use tissue binding capacity based on the relative ratio of \(B_m\) and \(K_d\) to explain the relative differences in tissue uptake that can be observed in balloon treated arteries. While each site in the tissue binds drug based on the magnitude of the affinity, the ability of the tissue as a whole to bind and retain drug is determined by the binding potential \((B_p = B_m/K_d)\), defined as the product of the binding capacity and the binding affinity. As binding potential increases, the fraction of drug that is bound by tissue proteins and retained also increases following drug delivery.\(^{5}\) Our experimental estimation of \(B_m\) for Zotarolimus in porcine femoral arteries was within the range of values predicted for Sirolimus in calf carotid arteries, whereas the estimate of \(K_d\) was almost

\(^{1}\) Note that \(\text{erfc}\left(\frac{L_p}{2\sqrt{D_t t/(1 + B_p)}}\right) = 0.083\) for \(t = 1\) hr.
10-fold higher than their counterparts.\textsuperscript{5} Taken together, these data imply that femoral arteries display a significantly lower binding potential for Zotarolimus ($B_p = 10.92$) than calf carotids for Sirolimus ($B_p = 140$).\textsuperscript{5} As Zotarolimus is more hydrophobic than Sirolimus, this difference in binding potentials partially explains the dependence between tissue type and total drug uptake. This inference is consistent with our observation that femoral arteries also displayed a relatively low binding potential for Paclitaxel ($B_p = 9.54$) compared to calf carotids ($B_p = 40$).\textsuperscript{5}

\section*{Study limitations}
Tissue incubation experiments were performed using excised porcine arterial specimens and we cannot discount the possibility that subtle structural changes might have occurred during the time between animal sacrifice and experiments. Such changes in tissue ultrastructure could impact estimates of drug binding and diffusivity. Nevertheless, the finding that only a small perturbation of the \textit{in vitro} measured diffusivity is required to match \textit{in vivo} tissue content (Figures 4A & 4B) suggests that such changes may be inconsequential and that bench-top experiments provide adequate estimates of \textit{in vivo} transport parameters.

Modeling of drug binding accounted for a unique type of binding site that was based on parameters estimated using bench-top experiments. These parameters are consistent with estimates relevant to low affinity binding to tissue proteins.\textsuperscript{7} Yet Zotarolimus also binds specifically and avidly with intra-cellular FKBP12. It is the net binding of Zotarolimus to both intra- and extra-cellular targets that leads to biological effects such as inhibition of neointimal smooth muscle cell proliferation and systemic immunosuppression.\textsuperscript{8} The net cellular and tissue level representations of the pharmacologic constants may provide a more accurate correlation between the model derived predictions and the \textit{in vivo} estimates of arterial drug uptake.\textsuperscript{7}
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Table S1: Two-sample Mann-Whitney test performed to compare the effect of balloon exposure time (30 s versus 180 s) on tissue uptake for SFA and iliac arteries.

<table>
<thead>
<tr>
<th>Incubation time</th>
<th>Percent difference in tissue uptake</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>5 min</td>
<td>94</td>
<td>0.014</td>
</tr>
<tr>
<td>4 hrs</td>
<td>74</td>
<td>0.078</td>
</tr>
<tr>
<td>24 hrs</td>
<td>40</td>
<td>0.034</td>
</tr>
</tbody>
</table>

Table S2: Two-sample Mann-Whitney test performed to compare 5x40 mm and 6x40 mm data for respective inflation and incubation times identifies no effect of balloon size on in vivo tissue uptake. Of the 72 balloons deployed, data from 2 balloons were not recorded, and data from 12 balloons were recorded without noting the balloon size (either 5x40 mm or 6x40 mm), leaving 58 balloons for this analysis. Those for which balloon size was not recorded were however included in subsequent analyses.
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Figure S1: Design of balloon inflations in iliofemoral arteries per animal. In a random fashion in each animal with distal treatment carried out first, arteries were exposed to 30±1 s or 180±2 s inflations.

Figure S2: Average tissue concentration (30 s balloon inflation case) showing the effect of zero concentration boundary condition defined by Equation 8.
Figure S3: Average tissue concentration (180 s balloon inflation case) showing the effect of zero concentration boundary condition defined by Equation 8.
**Figure S4:** Computational predictions of Zotarolimus arterial distribution after exposure to 180 s balloon inflation at two different time points: (A) and (B) at the end of balloon inflation, (C) and (D) at 5 min from the beginning of the procedure.

**References**