

# Multi-Drug-Loaded Microcapsules with Controlled Release for Management of Parkinson's Disease

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 $\boldsymbol{P}$ arkinson's disease (PD) is a progressive disease of the nervous system, and is currently managed through commercial tablets that do not sufficiently enable controlled, sustained release capabilities. It is hypothesized that a drug delivery system that provides controlled and sustained release of PD drugs would afford better management of PD. Hollow microcapsules composed of poly-L-lactide (PLLA) and poly (caprolactone) (PCL) are prepared through a modified double-emulsion technique. They are loaded with three PD drugs, i.e., levodopa (LD), carbidopa (CD), and entacapone (ENT), at a ratio of 4:1:8, similar to commercial PD tablets. LD and CD are localized in both the hollow cavity and PLLA/PCL shell, while ENT is localized in the PLLA/PCL shell. Release kinetics of hydrophobic ENT is observed to be relatively slow as compared to the other hydrophilic drugs. It is further hypothesized that encapsulating ENT into PCL as a surface coating onto these microcapsules can aid in accelerating its release. Now, these spray-coated hollow microcapsules exhibit similar release kinetics, according to Higuchi's rate, for all three drugs. The results suggest that multiple drug encapsulation of LD, CD, and ENT in gastric floating microcapsules could be further developed for in vivo evaluation for the management of PD.

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1. Introduction

Parkinson's disease (PD) is well-known as a progressive and degenerative disease of the nervous system. It is more common in the elderly, with most cases occurring after the age of 50. The degeneration of dopaminergic neurons in the substantia nigra, and a reduction in the amount of the neurotransmitter dopamine available in the striatum relate symptoms of this disease.<sup>[1–5]</sup> PD is currently treated through (i) dopamine substitution via administration of its metabolic precursor levodopa (LD), (ii) administration of drugs that inhibit the metabolism of dopamine, and (iii) activation of striatal dopamine receptors by dopamine agonists.<sup>[6,7]</sup> Dopamine substitution via LD is currently the "gold standard," as unlike dopamine, LD is able to cross the blood brain barrier. In a case of oral administration, however, LD undesirably undergoes extensive decarboxylation during its transport through the liver-a problem that requires it to be administered in combination with other drugs to improve its efficacy.



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LD is well-known as the most effective therapeutic agent for the treatment of PD.<sup>[8-10]</sup> LD is absorbed mainly in the proximal small intestine and then quickly metabolized to dopamine by the aromatic L-amino acid decarboxylase (AADC) enzyme. Its elimination half-life is approximately 90 min, and only a small amount ( $\approx$ 1%) of LD eventually reaches the brain after a single oral dose.<sup>[11]</sup> In addition, chronic administration of LD leads to LD-induced dyskinesia (LID). Dyskinesia occurs most commonly at the time of peak LD plasma concentrations during intermittent or pulsatile LD stimulation, and is referred to as peak-dose dyskinesia.<sup>[11–13]</sup> Achieving a continuous and sustained delivery of LD could nonetheless reduce the emergence of LID.

Carbidopa (CD) is in a class of medications called decarboxylase inhibitors. It works by preventing LD from being broken down before it reaches the brain. Hence, LD products that are formulated with CD as AADC inhibitor inhibit metabolism of LD at periphery and increase the number of fractions delivered to the brain for dopamine conversion.<sup>[14,15]</sup> In addition, LD can also be administered with a dopa-decarboxylase inhibitor that shifts the peripheral metabolism of LD to an alternative pathway of catabolismmethylation [catechol-O-methyltransferase (COMT)].<sup>[16,17]</sup> COMT inhibitors are able to block peripheral LD metabolism, prolong the half-life of LD and increase its delivery to the brain.<sup>[18]</sup> Entacapone (ENT) is one peripheral COMT inhibitor that extends the half-life of LD elimination by up to 85%.<sup>[19]</sup> The addition of ENT to LD in patients with moderate/severe PD achieved sustained plasma levels of LD, and therefore provides a more constant dopaminergic stimulation in the brain. This reduces the required dosage and frequency of LD administration, thus decreasing the "off" time and increasing the "on" time, which essentially results in an improved and prolonged clinical response to LD.[20-22]

Despite the synergism of these drugs in managing PD, their short elimination half-lives ( $T_{1/2}$ -LD = 50–90 min;  $T_{1/2}$ -CD = 1–2 h;  $T_{1/2}$  ENT = 0.4–0.7 h)<sup>[23]</sup> pose other challenges. For example, in one pharmacokinetic study, LD/CD (LC) and LD/CD/ENT (LCE) tablets administered four times daily at 3.5 h intervals to mice still showed insufficiency in achieving a continuous delivery of LD.<sup>[24]</sup> In order to maintain adequate plasma drug levels over prolonged periods, frequent dosing is therefore required. This would inevitably result in sharp fluctuations in plasma drug level, giving rise to end-of-dose "wearing-off" symptoms. Compounding to this problem, a large number of these patients are elderly and probably already taking multiple pills for other ailments. These elderly patients are also more likely to overlook or "miss" their dose. High pill burden therefore decreases compliance to drug therapy, arising from the need to take a large quantity of these pills on a regular basis. All these factors, in combination, increase the possibility of side effects and medication errors for PD patients. These challenges therefore necessitate for controlled and sustained delivery of PD drugs in order to reduce dosing frequency, while improving therapeutic efficacy and patient compliance.

At present, there are only a few medications that are commercially available to patients, and these include Sinemet, Stalevo, and Rytary. Unfortunately, Sinemet and Stalevo are

non-sustained releasing medications, and do not address to the above issues. Although Rytary has sustained release capabilities, it contains only two of the three PD drugs, i.e., LC/CD. Our group previously reported on a multi-drugloaded microcapsule delivery system that showed great buoyancy in simulated gastric fluid, while providing controlled release of multiple cardiovascular drugs.<sup>[25]</sup> The floatability of the designed microcapsules enabled for a prolonged gastric residence time, to provide sustained release while improving drug bioavailability. It is hypothesized that this microcapsule delivery system would provide controlled, and a more sustained, release of PD drugs as compared to commercially available PD tablets, thus potentially improving treatment outcomes and patient compliance. The aim of this study was therefore to develop a PD-specific drug delivery system that provides complete and simultaneous release of three PD drugs, i.e., LD, CD (both hydrophilic) and ENT (hydrophobic), towards better management of PD. Here, FDAapproved, biocompatible and biodegradable polymers, such as poly(L-lactide) (PLLA) and poly(caprolactone) (PCL), were used in the design of the delivery system.<sup>[26,27]</sup> Other design parameters, such as PLLA/PCL ratio, spray-coating and the amount and localization of drugs, to modulate drug release profiles were also investigated. Drug localization was examined through scanning electron microscopy (SEM), Raman mapping, and confocal laser microscopy. Drug release profiles from various formulations were evaluated in simulated gastric and intestinal fluid for up to 24 h. Subsequently, buoyancy and in vitro hydrolytic degradation of these microcapsules were also investigated to lend support to the drug release profiles and kinetics observed.

### 2. Results and Discussion

#### 2.1. Uncoated Microcapsules

Microcapsules co-loaded with LD, CD, and ENT were fabricated through a double emulsion  $(W_1/O/W_2)$  solvent evaporation method. In order to attain prolonged gastric residence time of these microcapsules, they were designed to be hollow, i.e., of lower density, to attain better floating properties. Modifications were therefore made to our previously reported fabrication process to produce hollow microcapsules through a single-step process. With the use of a rotary evaporator under reduced pressure, microcapsules with hollow cavities were obtained because of the fast solvent extraction rate under this condition (Figure 1). A fast solvent extraction rate shortens the time for any coalescence of the inner aqueous phase (W1) with the outer aqueous phase (W2), such that hollow microcapsules can be generated.<sup>[26]</sup> To achieve different drug release kinetics, several hollow microcapsule samples (i.e., F1 to F3) were fabricated (details are summarized in Table 1). While all samples were composed of PLLA/ PCL blends, they differed in their polymer blend ratios. The same amount of PD drugs were also encapsulated in these samples at a ratio of 4:1:8 (LD:CD:ENT), so as to replicate the same drug ratio as commercially available PD tablets (i.e., Stalevo-100) for comparison purposes.





Figure 1. SEM image of a cross-sectioned uncoated and spray-coated microcapsules loaded LD, CD, and ENT in SGF after 0 (left column), 6 (middle column), and 24 h (right column). Row 1: F1; Row 2: F2; Row 3: F3; Row 4: F4; Row 5: F5, Row 6: F6.

Figure 1 shows the SEM images of the fabricated microcapsules, and their corresponding images (from top to bottom) with increasing drug release time (left to right column). Regardless of the polymer blend ratios, all microcapsules are spherical in shape, have a mean diameter of  $\approx 600 \,\mu$ m, and possess a smooth surface finishing, uniform layer thickness and a hollow cavity. Unlike samples F1 and F2, sample F3 was, however, observed to have distinctively larger PCL particulates dispersed within the PLLA shell.<sup>[28]</sup> This was due to the phase separation of immiscible PCL within PLLA that resulted in such a structural morphology. With the exception of particle morphology, there were no other significant observable differences, across all three

samples. The drug loading (DL) (%) of LD, CD, and ENT in all these microcapsules were also comparable (Table S1, Supporting Information).

In order to increase drug bioavailability, these microcapsules should possess good buoyancy, so as to prolong their gastric retention time and allow for drugs to be released in the upper gastrointestinal tract. For example, it was reported that floating tablets exhibited higher bioavailability (162.7%) of Norfloxacin in healthy male human volunteers as compared to conventional non-floatable tablet forms.<sup>[29]</sup> In addition, it was also reported that propranolol HCl loaded into a floating gum showed a significant improvement in its bioavailability (132%) as compared with commercial propranolol



Table 1. Compositions used to achieve different release profiles of the microparticles.

Samples	Organic phase	Coating layer 1		Coating layer 2		Coating layer 3	
	Solution	Solution	Flow rate [mL min <sup>-1</sup> ]	Solution	Flow rate [mL min <sup>-1</sup> ]	Solution	Flow rate [mL min <sup>-1</sup> ]
F1-PLLA/PCL 3:1	ENT (0.08 g)	_	-	-	-	-	-
F2-PLLA/PCL 2:1	ENT (0.08 g)	_	-	-	-	-	-
F3-PLLA/PCL 1:1	ENT(0.04 g)	_	-	_	-	-	-
F4-PLLA/PCL 3:1 + PCL	ENT (0.04 g)	DCM (0.5 mL)	0.5	PCL (0.1 g) ENT (0.02 g)	0.3	PCL(0.1 g) ENT (0.02 g)	0.3
F5-PLLA/PCL 2:1 + PCL	ENT (0.04 g)	DCM (0.5 mL)	0.5	PCL (0.1 g) ENT (0.02 g)	0.3	PCL (0.1 g) ENT (0.02 g)	0.3
F6-PLLA/PCL 2:1+ PCL B	ENT (0.02 g)	DCM (0.5 mL)	0.5	PCL(0.1 g) ENT (0.03 g)	0.3	PCL (0.1 g) ENT (0.03 g)	0.3

tablets (Ciplar LA 80).<sup>[30]</sup> Buoyancy of the microcapsules was subsequently evaluated to examine their floatability in simulated gastric fluid (SGF) over 24 h (Figure 2). It was observed that microcapsule samples F1 to F3 possess floating capabilities, but to varying degree of success. The ability to stay afloat was attributed to their low density aided by an oilfilled hollow cavity, together with the hydrophobic nature of a PCL/PLLA blend that restricted rapid water uptake into these hollow microcapsules.<sup>[31-33]</sup> Sample F1 had 94% of the microcapsules remaining afloat at 24 h, whereas samples F2 and F3 fared poorer in this regards. Specifically, sample F3 only had  $\approx 60\%$  of the microcapsules afloat at 24 h. The floating tendency of these microcapsules was observed to be affected by its PCL/PLLA blend ratio, whereby floatability decreased with increasing PCL content. PCL, as an amorphous polymer with low glass transition temperature (below 0 °C), promoted the influx of water into the microcapsules (Figure 3a) The more open and mobile polymer chains in PCL resulted in higher water uptake,<sup>[3]</sup> thereby decreasing their buoyancy over time. This further resulted in the formation of pores that further promote water ingress, thus translating to inferior buoyancy.[34]

To test the hypothesis that this drug delivery system can better provide sustained and controlled release of triple PD drugs, as opposed to commercially available tablets, the release profiles of all three PD drugs from the samples were subsequently investigated. As observed from **Figure 4**a, commercial PD tablets (Stalevo) in SGF gave a complete release



**Figure 2.** Buoyancy (%) of the various microcapsules encapsulating three drugs in SGF at 37 °C for 24 h (n = 3).

of all three drugs within a short span of 2 h, as these tablets were not controlled-release formulations. Figure 4 further plots the drug release profiles for all three samples for each of the three PD drugs. A correlation between drug release kinetics and PCL content can be drawn from these plots, whereby all three drugs had faster release rates in samples with higher PCL content (i.e., F3 > F2 > F1). This, again, was attributed to the higher water uptake arising from the higher PCL content in the polymer blends.

The different release kinetics of the samples could be further explained by the hydrolytic degradation of these microcapsules (Figure 3). The faster release rate also correlated to the faster degradation of the PLLA in the PCL/PLLA shell capsule arising from the higher water uptake (Figure 3a). As PLLA has a semi-crystalline morphology and relatively more hydrophobic nature, PLLA retards any appreciable water influx,<sup>[35]</sup> which can explain for a slower release rate from sample F1. Microcapsule shells with a higher PCL content were more rubbery, due to the presence of larger phaseseparated PCL particulates (Figure 1). At 24 h, samples F3 were observed to have a more porous shell that would inevitably further accelerate drug release (Figure 4). Faster release rates were therefore attributed to larger PCL particulates that promote water ingress that increases the rate of PLLA degradation (Figure 3b,c). Taken in totality, these factors allowed for a faster diffusion of drugs through a more porous shell, translating to a faster release rate.

While the release rates for hydrophilic LD and CD were observed to be reasonable, and somewhat

observed to be reasonable, and somewhat related to a diffusion-controlled releasing mechanism, the release kinetics for the more hydrophobic ENT was observed to be relatively much slower. In order to replicate the drug content from commercial tablets at a ratio of 4:1:8 (LD:CD:ENT), it was essential to deliver all three drugs at the same rate. From Figure 4d, the release of hydrophobic ENT from F1, F2, and F3 was found to be highly retarded at only  $\approx$ 13.2%, 25.3%, and 42.5% within 24 h. As the partition coefficient of ENT for [octanol]/[water] is 2.0,<sup>[36]</sup> the slow release of hydrophobic ENT is expected. At the same time, a more hydrophobic ENT



**Figure 3.** a) Water uptake, b) change in molecular weight of PLLA, and c) mass loss of the degrading uncoated microcapsules as a function of incubation time (n = 3).

would interact strongly with PLLA through hydrophobic– hydrophobic affinity, further retarding its release.<sup>[37]</sup>

#### 2.2. Spray-Coated Microcapsules

Although the hollow microcapsules showed some promise, i.e., good buoyancy and the ability to sustain the release of LD and CD, the release of hydrophobic ENT was too slow and did not match up to the release kinetics of the other two hydrophilic drugs. To address to this issue, it was further hypothesized that the encapsulation of ENT into PCL as an additional surface coating onto the hollow microcapsules can aid in accelerating its release, by reducing the drug diffusion distance to the release environment. To test this hypothesis, samples F1 and F2 were chosen for further development, by coating with PCL–ENT; but with differing



ENT amounts (Table 1). In order to keep to the same drug ratio of 4:1:8, half of the original amount of ENT was loaded into the hollow microcapsule, while the other half was coated onto the surface of the microcapsule, i.e., PLLA/PCL 3:1 (F4) and PLLA/PCL 2:1 (F5). A third sample, F6, had a quarter of the original ENT amount loaded in the hollow microcapsules, while the remaining ENT was surface coated. The rationale of choosing PCL as the coating polymer was because of its amorphous nature that allows for a faster water uptake which is expected to translate to a more rapid release rate.

Figure 1 displays the SEM images of samples F4 to F6. It was observed that freshly spray-coated microcapsules (t = 0 h) were likewise spherical in shape, but had an additional coating layer on the surface. With the additional coating, the size of the microcapsules was observed to increase by about 100 µm. To determine the drug and polymer localization within the microcapsules, Raman spectroscopy and mapping (Figure 5) was conducted for sample F5, as a representative sample. Figure 5a shows the Raman spectrum of each drug and polymer in the microcapsule, from the inner hollow cavity to the particle shell and finally to the outer coating layer. Raman mapping was further employed to distinguish the distribution of the polymers and drugs. Raman mapping shows that while the hydrophobic ENT had preferential interactions with the hydrophobic polymer shell, the hydrophilic LD and CD drugs were uniformly distributed both in the hollow cavity and capsule shell. As ENT itself possesses yellow fluorescence, confocal laser scanning microscopy (CLSM)

was utilized to determine its distribution. CLSM images obtained for F5 are shown in **Figure 6**. The images further confirmed the localization of ENT both within the microcapsules and on the surface of the microcapsule, as a uniform coating layer. From the z-stack images, the surface of the particle exhibited the strongest fluorescence from ENT, confirming the effectiveness of the spray coating process. The largest amount of ENT was localized in the outer layer, i.e., coating/shell, while the hollow core exhibited a much weaker intensity.

As seen in Figure 5c, it was also confirmed that the spraycoated microcapsule had a hollow cavity, a shell and a coating layer. Both Raman mapping showed strong intensity of PLLA and PCL in the shell and coating layer, respectively. A relatively weaker intensity of PCL was observed in the capsule shell as compared to PLLA, as the larger phase would naturally engulf the minor phase (i.e., PCL). Interestingly, no







Figure 4. Release profiles of a) commercial tablets (Stalevo), b) LD, c) CD, and d) ENT from uncoated microcapsules in SGF at 37 °C for 24 h (n = 3).

significant intensity for PLLA and PCL was obtained in the hollow cavity of the microcapsule. On the contrary, a uniform distribution of the hydrophilic drugs was observed here. This phenomenon could be explained as during rapid evaporation of organic solvent (DCM) using rotary evaporator, the hydrophilic LD and CD in the water phase remained in the hollow cavity, likely as an internal wall coating, while the more hydrophobic PLLA/PCL shell solidified with ENT-embedded in this shell. Notably, the spray-coated PCL layer on uncoated microcapsules exhibited no affect to their buoyancy.<sup>[25]</sup> The F4, F5, and F6 exhibited 90 ± 6, 85 ± 6, and 82 ± 7% of buoyancy after 24 h.

The release profiles of each drug from PLLA/PCL microcapsules with PCL coating layer in SGF are shown in **Figure 7**. PCL coated microcapsules (F4–F6) exhibited significantly faster release of ENT compared to that from uncoated

microcapsules (F1–F3). Furthermore, when a higher amount of ENT was encapsulated in PCL coating layer (F6), the release rate and cumulative release of ENT were faster and higher in quantity, respectively (Figure 7c).

As it had been reported that floating microspheres are able to provide prolonged GRT of 5 h in the human body,<sup>[38,39]</sup> release study was further conducted but, this time, in an environment that would simulate drug release in the stomach, based on typical gastric emptying time, followed by release in the intestinal region<sup>[39–42]</sup> As such, the release profiles of these microcapsules (F4, F5, and F6) were conducted in SGF, followed by simulated intestinal fluid (SIF) (**Figure 8**). Since the major absorption site of all three drugs was reported to be the stomach and upper intestine,<sup>[43–46]</sup> having a 70% release of all three drugs in this regions within 5–7 h would definitely be advantageous. From Figure 8, all



Figure 5. a) Raman spectra of pure component, b) spectra from hollow core, shell, and coating layer of F5 and b) Raman mapping images of F5.





Figure 6. CLSM images of a cross-sectional microcapsule (F5). a) Emission of the ENT, b) no laser, c) emission of the ENT, and d) z-stack images from center to top of microcapsule.

the samples now exhibited relatively similar release rates for all three drugs in SGF. Here, the ENT-loaded PCL coating layer and the ENT-loaded capsule shell provided an initial and sustained release of ENT, respectively. The ENT-loaded rubbery PCL coating layer onto microcapsule not only shortens drug diffusion distance, but also accelerated drug dissolution and diffusion. With PCL as a coating layer, the degradation of PLLA shell was observed to be retarded as compared to the non-coated microcapsules (Figure S1, Supporting Information).

The new formulations were also able to achieve sustained delivery of all three PD drugs, releasing up to 80% of the

drugs within 24 h. This new coated formulation could modify release profile of multiple drugs regardless of their hydrophilicities by altering their PLLA/PCL ratio and through the use of an additional coating technique. To elucidate the mechanism of drug release from these coated microcapsules, the release data were fitted to zero order kinetics (cumulative amount of drug released vs time),<sup>[47]</sup> first-order kinetics (log cumulative percentage of drug remaining vs time),<sup>[48]</sup> and Higuchi's release kinetics (cumulative percentage of drug released vs square root of time).<sup>[49]</sup> **Table 2** summarizes the  $r^2$  correlation values for all three fits and the corresponding release rate constants. From the data, Higuchi's equation has



Figure 7. In vitro release profiles of a) LD, b) CD, and c) ENT from coated microcapsules in SGF for 24 h at 37 °C (n = 3).

the best fit of all. Sample F4 had the lowest k values for all drugs as compared to F5 and F6 because of the PLLA/PCL ratio effects that were earlier discussed. Among the three samples, F6 had the closest k values for all three drugs. The total amount of LD, CD, and ENT in the commercial tablet (Stalevo-100) is 100, 25, and 200 mg, respectively. As F6, with a drug ratio of 4:1:8 (Table S1, supporting information)

similar to commercial tablets, had the closest k values for all three drugs (Table 2), F6 could potentially be exploited for management of PD in patients. This would be further evaluated through in vivo pharmacodynamics studies. In terms of the weight of F6 required in patients, this would correspond to approximately 1.82 g of this microcapsule to meet the same amount of drugs as the commercial tablet—Stalevo-100.



Figure 8. In vitro release profiles of LD, CD, and ENT from a) F4, b) F5, and c) F6 in SGF for 5 h followed by release into SIF for 24 h at 37 °C (n = 3).

Table 2.	Kinetics	of LD,	CD ar	d EN1	from	coated	microcapsules.
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Drug	Formulation		Rate constant $k$ (h <sup>-1/2</sup> )		
		Zero-order	First-order	Higuchi equation	Higuchi equation
LD	F4	0.7145	0.8694	0.9472	15.023
	F5	0.6436	0.8596	0.8826	17.245
	F6	0.7102	0.8717	0.9315	17.735
CD	F4	0.6923	0.8966	0.9526	15.815
	F5	0.704	0.8665	0.9150	17.052
	F6	0.752	0.9067	0.9310	17.345
ENT	F4	0.7214	0.8346	0.9290	12.249
	F5	0.6387	0.7865	0.8816	15.488
	F6	0.5305	0.7089	0.8032	16.615

Our spray-coated microcapsule may therefore be a promising oral drug delivery system to achieve sustained and controlled release of multiple drugs for the management of Parkinson's disease.

### 3. Conclusion

Floatable microcapsules loaded with three different PD drugs, and with the same drug ratio as commercial PD tablets, were fabricated. Manipulating polymer ratio between PLLA and PCL can influence how these drugs were released. However, because of the hydrophobic nature of ENT, its release was greatly retarded. Coating of these microcapsules with ENT and PCL onto its shell was therefore introduced as a technique to load a portion of ENT onto the surface of the microcapsules. With this, the release profiles fitted well into the Higuchi's release kinetics and sample F6 was found to have similar release kinetics ( $k = 17.23 \text{ h}^{-1/2}$ ) for all three PD drugs. This delivery system would be further evaluated through in vivo studies in comparisons against commercially available PD tablets.

### 4. Experimental Section

*Materials*: PLLA (IV: 2.4, Purac), PCL (molecular weight 10 kDa, Sigma-Aldrich), and Polyvinyl alcohol (PVA) (molecular weight 30–70 kDa, Sigma-Aldrich) were used without further purification. LD, CD, ENT, Tween 20, HCl solution (37% v/v Fuming), acetic acid, and ethyl acetate (ETA) were purchased from Sigma-Aldrich (Steinheim, Switzerland). Dichloromethane (DCM), acetonitrile, chloroform, and tetrahydrofuran (THF) were purchased from Tedia Co. Inc. Olive oil (Pietro Coricelli) was used. All other chemicals and reagents used were of analytical grade. The SGF (pH 1) was prepared by adding 0.1  $\bowtie$  HCl solution to 0.02% (w/v) Tween 20. The SIF was prepared by mixing pH 6.8 phosphate buffer and 0.02% (w/v) Tween 80.

Preparation of Multi-Drug-Loaded Hollow Microcapsules: Encapsulation of free LD and CD in PLLA/PCL polymer mixture was performed using the water–oil–water  $(W_1/O/W_2)$  double emulsion



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method. The internal aqueous solution was prepared by adding 1 mL of 0.5  $\,$  M HCL to 50.0 mg of LD, 12.5 mg of CD and vortex until fully dissolved. The resultant solution was then introduced dropwise above the polymer solution (0.3 g of PLLA and 0.1 g of PCL dissolved in 5 mL of DCM) and then 10  $\mu$ L of olive oil was added under magnetic stirring. The obtained mixture was added to 0.25% (w/v) PVA solution (50 mL) containing DCM (1 mL) and stirred using overhead stirring at 250 rpm for 4 min. For preparation of mixture of PVA solution (50 mL) containing DCM (1 mL), DCM (1 mL) was added into 0.25% (w/v) PVA solution (50 mL) containing trip expression of mixture of PVA solution uniformly. Since DCM is not miscible in water, DCM was dispersed as small droplets.

The resultant emulsion was quickly transferred to rotary evaporator with the addition of 0.25% (w/v) PVA solution (150 mL) to solidify the microcapsules. The microcapsules obtained were centrifuged, washed with distilled water for three times, freeze dried, and kept in a desiccator.

Spray-Coating on Microcapsules: The spray-coating was performed using MediCoat DES 1000 Benchtop Stent Coating System (Sono-TekCorp., Milton, NY). The coating process was conducted in a row-by-row sweeping manner. The settings were identical and included power for the frequency generator of 1.0 W for coating layer 1, 1.4 W for the remaining layers and an air shroud level of 2 psi. Table 1 shows each layer's spraying composition. Finally, the spray-coated microcapsules were dried in a vacuum oven at 40 °C for 24 h.

*Drug Loading Content*: Microcapsules (10 mg) were accurately weighed and dissolved in 1 mL of DCM through sonication. SGF (10 mL) was then added, and mixed using a vortex at 300 rpm (n = 3). The hydrophilic LD and CD partitions into SGF and the supernatant was analyzed using reverse phase high performance liquid chromatography (RP-HPLC) with 100% acetic Acid (2% (v/v)) as mobile phase at wavelength 284 nm. The ENT is then re-dissolved in sodium dihydrogen phosphate (60%)/methanol (40%) mixture to precipitate polymer. The supernatant is then taken and filtered through a 0.22 µm syringe filter. The resultant solution is then analyzed using RP-HPLC with sodium dihydrogen phosphate (60%)/methanol (40%) mixture as mobile phase at wavelength 284 nm. The following equation was applied to calculate the DL. DL (%) = weight of the drug in microcapsules × 100%/weight of the microcapsules.

*Morphological Analysis*: The cross-sectional morphology of microcapsules prepared was observed with a SEM (JEOL JSM-6360A). First, the microcapsules were mounted on carbon tape attached metal stub and cross-sectioned using a surgical blade to expose the internal cross sections and the microcapsules were subsequently coated with gold using a sputter coater (SPI-Module). To measure average capsule size, The Image J software was used to measure capsule size.

Raman Mapping: Raman mapping was used for observing the polymer and drugs distribution within the microcapsules. Cross-sectioned microcapsule was put under a microscope objective with a laser power of 10 mW. Raman mapping measurements were carried out with a step size interval of 5  $\mu$ m to from a grid map using a Raman microscope (Nicolet is50, Thermo Scientific) equipped with a near-infrared enhanced deep depleted thermo-electrically Peltier-cooled CCD array detector and a high-grade Leica microscope. The pre-sectioned microcapsule was irradiated



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with a 785 nm near-infrared diode laser, and the back scattered light was collected by an objective lens. Measurement scans were collected in a spectrum range from 200 to 3200 cm<sup>-1</sup>.

Confocal Laser Microscopy (CLMS): ENT distribution within microcapsule was analyzed with a CLSM (LSM710). The aqueous suspension containing particles were drop-wise added to a glass slide before sealing with a glass cover slip. All photos were obtained using  $63\times/1.40$  oil objective lens, and the AxioCan MRm camera. Analysis of the images was done with the ZEN 2012 software.

Hydrolytic Degradation: Microcapsules were weighed (50 mg) and placed in glass bottles filled with SGF (20 mL). Samples were incubated 37 °C with gentle shaking. At pre-determined time points, microcapsules were collected from the bottles. For water uptake study, the microcapsules were washed with distilled water, weighed, and dried to obtain the dry mass. The percentage of water uptake was calculated at pre-determined time point as the difference between the mass of the wet and dry microcapsules, measured at time *t*, and taken as a percentage of the dry weight. Each experiment was conducted in triplicate. To determine mass loss, microcapsule mass loss was calculated as the difference between the initial mass of the microcapsules and dry mass at time *t*, and calculated by dividing by the initial mass and dry mass. Finally, molecular weight of the microcapsules was measured using the Agilent GPC 1100 Series using a reflective index detector (RID) at 30 °C. Chloroform used as solvent and the flow rate was 1 mL min<sup>-1</sup>. Based on the solubility differences of the polymers in THF (PCL is soluble in THF, while PLLA is not), the two polymers in the microcapsules were separated by the dissolution method. Each microcapsule (10 mg) was added in THF (1 mL) to dissolve the PCL. The mixture solution was evaporated at room temperature for 48 h. The remaining solvent in the solution was further dried in an oven at 40 °C for a 48 h. And then, chloroform (1 mL) was added to dried PLLA and analyzed for GPC. Molecular weights of the microcapsules were calculated by the calibration curve using polystyrene standards (165-5000 kDa).

Buoyancy Test: A visual observation method was used to measure the buoyance of the various microcapsules.<sup>[38]</sup> The microcapsules were added into in the transparent glass vials filled with SGF (20 mL). And then, the vials were placed under stirring at 250 rpm at 37 °C for 24 h. At each time point, the number of microcapsules floating on the SGF was counted visually. Finally, the percentage of floating microcapsules was calculated according to the ratio of the number of microcapsules floating on the SGF to the number of microcapsules used.

*In Vitro Release Study*: Release study was conducted in both SGF and SIF for 24 h. Twenty milligram of microcapsules was added in SGF (20 mL) each. Both samples were placed in a 37 °C rotating incubator. The medium (10 mL) in each bottle was extracted and replaced with 10 mL new medium at different time points. The time points included intervals of hours within a 24 h period. The drug content was analyzed using RP-HPLC.

*Kinetics Release Profile*: In order to analyze the kinetics of drug release, the release data were fitted to Zero-order, First-order and Higuchi equations.<sup>[47–49]</sup>

Statistical Analysis: Student's *t*-test was used to compare the groups. Statistically significant differences were considered when p value < 0.05. All data are expressed as the mean  $\pm$  standard deviation from three independent experiments.

## Supporting Information

Supporting Information is available from the Wiley Online Library or from the author.

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