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Release of antibiotics from electrospun bicomponent fibers

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Abstract Biocompatible nanofibers that are capable of adapting to the physiological conditions of the human body have become increasingly important for clinical applications in recent years. Electrospun fiber mats offer particular advantages due to their large surface area and their sorption/release properties. If loaded with drugs, delivery properties can be tailored to a specific release rate. This research work focuses on poly(L-lactic acid) (PLA) and poly(ε-caprolactone) (PCL) incorporating three different model antibiotics as well as bicomponent fibers made from PLA and PCL containing the same model drugs. Tetracycline and chlorotetracycline hydrochloride, and amphotericin B were selected as model drugs and their release properties and antimicrobial effectiveness studied. The surface morphology and the average diameter of the fibers strongly depended on the individual spinning system which in turn influenced the release of the therapeutic compounds from the fibers. Tetracycline was discharged from PCL at the highest rate while amphotericin B was slowest. PCL almost completely liberated any of the drugs over time while PLA only released about 10% total. By forming bicomponent PCL–PLA fibers surface and release characteristics could be modified to fit a sensible drug delivery.

Keywords Electrospinning · Biocompatible bicomponent fibers · Rheology · Poly(ε-caprolactone) · Poly(L-lactic acid) · Drug release

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Background

Electrospun fibers with diameters in the micro- to nanometer range have gained renewed interest over the last decade for their versatility in applications. Electrospinning is a fairly simple method to obtain fibrous webs with a large surface area to volume ratio. Numerous publications have dealt with the various aspects of the fiber formation process as well as the characteristics of the resulting micro- or nanofibers (Doshi and Reneker 1995; Reneker and Chun 1996; Smith and Ma 2004; Bognitzki et al. 2001; Deitzel et al. 2001; Lee et al. 2002; Huang et al. 2003; Gibson et al. 2001; Gibson 2003; Shin et al. 2001).

During the electrospinning process a polymer solution is formed into a ultra-fine fiber by applying an electric field between the spinneret (often a syringe) and a grounded collector device. The polymer jet is drawn to the collector by a whipping motion and deposited as a random web. Concerns have been raised that the fiber is of low orientation and crystallinity, thus has little tenacity. Since the

bending instability is not a static phenomenon, any form of drawing of the fibers is not uniform along the fiber axis. Applications for which tenacity is of minor significance, such as filtration devices, have been studied most intensively (Chung et al. 2002). Electrospun ultra-fine fibers are also well suited for engineered tissue replacement and therapeutic patches because of the flexibility and elasticity of the fibrous webs (Smith and Ma 2004; Yoshimoto et al. 2003). Beads formed as a consequence of incorrect solution properties and usually regarded as defects, can help to stabilize the web for these applications (Fong et al. 1999).

Biocompatible nanofibers that have been approved by the Food and Drug Administration (FDA) for medical applications are of specific interest (Chu 1997). Examples include tissue repair, wound healing devices and materials suitable for controlled drug delivery (Boland et al. 2001; Zong et al. 2002; Simpson et al. 2004; Kidoaki et al. 2004). For those materials porosity and morphology are important characteristics since adsorption/recognition processes occur at the interface of the biomaterial and the physiological environment. Published research mostly focused on the rate and control of drug release, and the chemical integrity of the drug within the polymer/solvent system (Kewany et al. 2002; Kenawy 2007; Kim et al. 2004). Incorporating an additional component into the polymer/solvent system, such as a drug, strongly affects solution properties. Viscosity, surface tension and conductivity may considerably change (Luong-Van et al. 2006). Further, Zeng et al. (2005) recognized that the location of the drug within the electrospun material largely determined its release characteristics. In their study poly(L-lactide) was spun from a mixture of acetone and chloroform and used as the carrier polymer for two drugs effective in cancer therapy. Release of water-soluble anti-inflammatory drugs from poly(vinyl alcohol) (PVOH) was investigated by Taepaiboon et al. (2006) and compared to delivery from cast PVOH films. The effect of the properties and concentration of the spinning solution on the morphology of the electrospun fibers and the chemical integrity of the drugs were taken into consideration. The fiber mats released more drugs in a shorter time than the respective films which is not surprising due to the lower surface area of films compared to nanofibrous webs. Similar results were obtained from cellulose acetate electrospun fibers and cast films loaded with four different pain medications for medical patches (Tungprapa et al. 2007), and vitamins delivered to the skin (Taepaiboon et al. 2007).

Electrospun bicomponent fibers have additional advantages, such as differential drug release properties, added surface due to different dissolution speeds under biological conditions, different morphologies, etc. If the fibers are electrospun from melt, they are usually larger in diameter (Hawkins et al. 2004; Buschle-Diller and Erdonmez 2003) than if spun from solvent solution, however their production is easier to handle. For electrospinning of bicomponent fibers from solution, not only the experimental electrospin geometry (e.g., the spinneret arrangement) is decisive, but also the properties of the spinning solution play a very important role. Parameters include the type and concentration of polymers and their inherent characteristics, solvents for each component, their miscibility, compatibility and evaporation rate. Interrelated with the solution viscosity and the shear properties are polymer molecular weight, degree of branching, degree of substitution, surface charge, etc. In case the bicomponent fibers are intended for therapeutic compound delivery, the properties of the drug add further complications. Both its chemical integrity and timely availability of sensible doses have to be considered.

Detailed studies on the formation of bicomponent fibers and the relationship of various solution properties during electrospinning are available (Theron et al. 2004), although the product fibers were not necessarily all destined to be used in the medical field or loaded with active components. For example, Kahol and Pinto (2004) studied the extrusion of polyaniline and poly(ethylene oxide) from chloroform. Polyamides, dissolved in formic acid, have been electrospun together with poly(ethylene terephthalate) from trifluoro acetic acid (Kit and Jagannathan 2002). Gupta and Wilkes (2003) reported work on the challenge to electrospin side-by-side bicomponent fibers made from poly(vinyl chloride) and segmented polyurethane as well as of poly(vinyl chloride)/poly(vinylidiene fluoride) dissolved in N,N-dimethyl acetamide. Jiang et al. (2005) managed to produce coaxial electrospun fibers from poly(ε-caprolactone) (PCL) and polyethylene glycol (PEG) containing proteins for release. Bicomponent

scaffolds for tissue repair were created by co-polymerization of poly(lactide-co-glycolide) (PLGA) random copolymer and a poly(DL-lactide)–poly(ethylene glycol) block co-polymer (Luu et al. 2003). Chitosan, a difficult-to-electrospin natural polymer, has been combined with poly(ethylene oxide) (PEO) and bicomponent fibers could be produced at low ratios of chitosan-to-PEO (Spasova et al. 2004). Li and Hsieh (2005) examined the various properties of solutions of poly(acrylic acid), their interrelationship and their effect on the properties of the product fibers.

The goal of this research was to electrospin fiber mats from PCL and PLA single and bicomponent polymers carrying different therapeutic compounds with potential use for medical applications. The discharge of the drug into a physiological buffer was monitored over a defined length of time and its antimicrobial effectiveness tested against *Escherichia coli* and *Staphylococcus aureus*. It was investigated how the choice of solvent and composition of the polymer solution determined the surface morphology and release properties of the product fibers.

Experimental procedures

Materials

Poly(ε-caprolactone) (PCL) and poly(L-lactic acid) (PLA) were obtained from Sigma-Aldrich. Chloroform and other solvents were analytical grade (Fisher Scientific, Atlanta, GA). Tris buffer (Trizma), chlorotetracycline hydrochloride (C; 76% purity), tetracycline hydrochloride (T; 95% purity) and Amphotericin B (AB; 80% purity) were obtained from Sigma-Aldrich (Atlanta, GA) and used as received.

Electrospinning conditions

Details of the electrospinning procedure are documented elsewhere (Buschle-Diller et al. 2006). Initially electrospinning experiments were conducted using chloroform, tetrahydrofuran (THF) or toluene as the solvent for both PCL and PLA as well as for the formation of bicomponent PCL–PLA fibers. Table 1 shows the composition of the spinning solutions and the flow rate. All solutions were mixed in vials and spun from an 18 Gauge needle at 20 kV with a fixed spinneret-target distance of 19.4 cm. As collector metal grids or scanning electron microscope stubs were used.

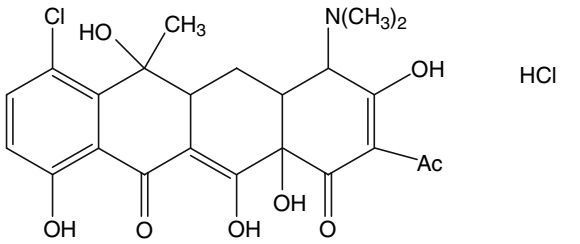
Table 1 Composition of polymer/copolymer spinning solutions and flow rates

Sample composition	Concentration [%]	Flow rate [mL/h]
PCL/no PLA	9	6
PCL/no PLA	12	10
PCL/no PLA	15	15
PCL/PLA, 3:1	12	10
PCL/PLA, 1:1	12	10
PCL/PLA, 1:3	12	10
PLA/no PCL	12	10

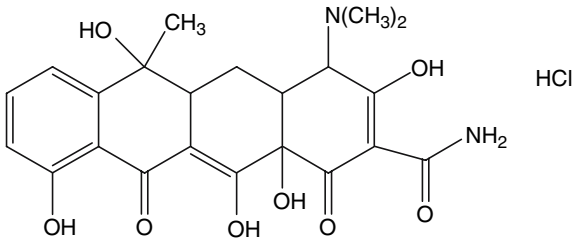
For the release studies 10 wt.% PCL in chloroform was mixed with 2 wt.% tetracycline hydrochloride (T) (based on weight of polymer); 2 wt.% chlorotetracycline (C); and 1 wt.% Amphotericin B (AB), respectively. T and C were predissolved in 1 mL methanol before adding to the spinning solution; AB was predissolved in 2 mL dimethylformamide (DMF). The solutions were stirred until homogeneous and electrospun at 20 kV with a tip-to-collector distance of 16 cm. The chemical structures of T and C are shown in Fig. 1. Amphotericin B ($C_{47}H_{73}NO_{17}$) is a polyene antibiotic from *Streptomyces* with antifungal properties. Bicomponent fibers containing PCL:PLA ratios of 3:1, 1:1 and 1:3, respectively, were spun from 12 wt.% chloroform, loaded with 2.15 wt.% T predissolved in 1 mL methanol. Electrospinning conditions were the same as those for unloaded bicomponent fibers (see above).

Assessment of solvent evaporation

A Mettler Toledo HB43 Halogen Moisture Analyzer was used to study the evaporation rate of the solvent. For each polymer solution 1.2 g were weighed into small glass tubes and tested for 10 min at 50, 60, 80 and 95 °C. Data were recorded by weight at every minute and drying curves constructed for each temperature setting. Tests were performed in duplicate.



Acetyl-2-decarboxamido-2-chloro-7-tetracycline (chlorotetracycline)



Tetracycline hydrochloride

Fig. 1 Chemical structures of model antibiotics C (chlorotetracycline hydrochloride) and T (tetracycline hydrochloride)

Determination of the properties of spinning solutions

Shear rate and viscosity were measured with a HAAKE RheoStress RS75. Polymer solution was placed between two plates with the upper plate being able to rotate with controlled rate based on shear rate. The viscosity of the polymer solution was determined by two test methods. Using the first method, the temperature was set, and shear rate changed from 0.15 to 300 s^{-1} within different set time periods: 60, 120, 180, 240, 300, 360, 420, and 480 s. With the second method, viscosity was tested at different temperatures: 23, 30, 40, and 50 °C. At each temperature setting, the shear rate was changed from 0.15 to 300 s^{-1} within 300 s.

Determination of surface tension of polymer solutions

The surface tension of the polymer solutions was tested with a Cahn Dynamic Contact Angle device (DCA 322). A 22 mm × 22 mm cover glass was fixed with a clip connected to a microbalance. A beaker with polymer solution was placed on a traveling platform which could be raised and lowered. The cover glass was immersed into polymer solution for 4 mm, and then returned to the original position at a controlled rate of 264 μm/s. During this process, the dynamic wetting forces on the cover glass were measured by an electronic microbalance.

Surface investigation by scanning electron microscopy (SEM)

Fibers were directly electrospun on SEM stubs covered with carbon tape and sputter-coated with gold for 2 min. A Zeiss DMS 940 scanning electron microscope was used at 10 kV to observe fiber surfaces at various magnifications.

Assessment of release properties and antimicrobial activities

The drug-loaded fiber mat was weighed and placed in 50 mL of 0.05 mol/L Tris buffer adjusted to pH 7.35, incubated at 37 °C and shaken at 100 rpm. The concentration of released drug was determined with a UV–vis spectrophotometer at time intervals of 2 min up to 3 h or longer depending on the sample type. The experiments were performed in duplicate or triplicate with freshly prepared drug-loaded fiber mats for each set.

To test antimicrobial properties, 25 mL bacterial suspension, containing either *Staphylococcus aureus* or *Escherichia coli*, was placed between two pieces of electrospun PCL fiber mats (25 mm × 25 mm) and held in place by a sterile weight. The bacterial suspensions contained 10^6–10^7 colony forming units (CFU). The actual number was determined by counting after spread-plating on trypticase soy agar plates. After 10, 30, and 60 min, the samples were placed in sterile conical centrifuge tubes, each containing 5.0 mL of sterile 0.01 M sodium thiosulfate and vortexed for 150 s to remove the bacteria. The plates were incubated at 37 °C for 24 h and then counted for viable CFU of bacteria. The reduction in bacterial count was calculated according to: reduction (%) = (B − A)/B × 100, where A and B are the CFU for the plates containing test samples and the control, respectively.

Results and discussion

Formation of fibrous PCL and PLA mats containing antibiotics

Important parameters for electrospinning are viscosity, surface tension, film-forming capabilities and factors introduced by the solvent (Luong-Van et al. 2006). Active compounds, such as drugs, intended for discharge from the product fibers further alter the properties of the system. While keeping the experimental set-up constant, such as electrode distance and voltage, fiber forming conditions were established based on the choice of solvent, the polymer concentration and the related behavior of the resultant fibers. A suitable solvent should be able to dissolve both PLA and PCL equally well, and electrospinning of single-component as well as bicomponent fibers with/without contained drugs should be possible.

For both PLA and PCL, solvents such as toluene, THF and chloroform were possible choices with polymer concentrations between 9 and 15 wt.%. Surface tension and viscosity of spinning solutions can be determining factors for the feasibility of the electrospinning process of a particular polymer but neither parameter clearly predicts the probability of fiber formation. On the other hand, measuring the rate of solvent evaporation can give a very good indication (Hawkins and Buschle-Diller 2004; Buschle-Diller et al. 2005). Polymers difficult to electrospin, such as, for example, chitosan, show almost linear evaporation curves which point to a slow, more or less even drying of the polymer solution throughout the fiber formation process. Such solutions form brittle fibers or break up into beads during the spinning process. In contrast, polymers that were easily electrospun into fine fibers developed a fast drying skin around a still liquid core, and evaporation measured as weight loss dropped exponentially.

Table 2 shows the results of surface tension measurements for solutions of PCL and PLA in chloroform at different concentrations as well as of bicomponent PCL–PLA systems. As expected, the surface tension within a given system increased with the concentration. Overall lower concentrations yielded fibers with somewhat smaller diameters. The bicomponent PCL–PLA spinning solutions did however not differ much in surface tension irrespective of their composition, and fibers could be formed from all three solvents under the given conditions.

Table 2 Surface tension (dynes/cm) of poly(ε-caprolactone) (PCL) and poly(L-lactic acid) (PLA) in chloroform in relationship to their concentration (standard deviation in parenthesis)

Sample composition	Surface tension (dynes/cm)
9% PCL	30.65 (±0.30)
12% PCL	32.42 (±0.38)
15% PCL	33.49 (±0.42)
12% PLA	30.55 (±0.11)
12% PLA–PCL (1:1)	30.76 (±0.35)
12% PLA–PCL (3:1)	31.55 (±0.36)
12% PLA–PCL (1:3)	31.11 (±0.12)

Significant differences, however, were observed in the solvent evaporation behavior from the three spinning solutions (Fig. 2). Evaporation of toluene occurred at a very low, almost linear rate, independent of the evaporation temperature used, and indicates behavior observed with spinning solutions with delayed coagulation. The product fibers were fused at contact points. Clearly the necessary skin could not form and the electrospun products were unsatisfactory. Similar results were obtained with THF as a solvent. Chloroform, on the other hand, provided a suitable solvent for both PCL and PLA. The shape of the evaporation curves of both polymers

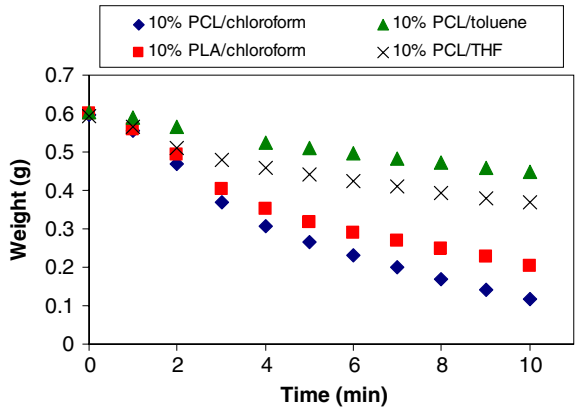


Fig. 2 Evaporation curves of 10 wt.% poly(ε-caprolactone) (PCL) in chloroform, toluene and tetrahydrofuran (THF), respectively, and poly(L-lactic acid) (PLA) in chloroform. Maximum standard deviation: ±0.02

not only suggested the fast formation of the essential outer film around the still liquid core, but also showed that the rate of solvent evaporation from both PCL and PLA solutions were very similar, implying that chloroform could very well be used as the solvent for bicomponent fiber formation.

The viscosity/shear rate determinations confirmed these observations. Time-dependent viscosity/shear rate measurements at a set temperature showed that the viscosity initially dropped sharply at a shear rate lower than 20 s^{-1}, then leveled off with only a small further decrease at higher shear rates. The consistent shape of the curves indicated the stability of the polymer solution under shear stress over time, a factor that is important for fiber formation via the electrospinning process. From these results it can be assumed that the formed polymer film is coherent and breaking of the polymer jet during the spinning process is not likely (Buschle-Diller et al. 2005).

Since PCL and PLA solutions in chloroform overall produced the most consistent product fibers, chloroform was used for all further experiments. With a concentration of 9 wt.%, PCL spinning solutions had the most suitable viscosity while concentrations up to 15 wt.% were still possible. PLA spinning solutions were more manageable at concentrations of 12–15 wt.%. In Fig. 3 a scanning electron micrograph (SEM) of a PCL fiber spun from 15% solution in chloroform is presented. The surface shows indentations of oval shape, varied in size and seemingly fairly deep. In comparison, the surface of electrospun PLA fibers appeared rougher, almost wrinkled with smaller round serrations (Fig. 4). PCL fibers varied in fiber diameter from very fine (300–400 nm) to rather large (6–7 μm) for a few fibers. At lower concentrations (9–10%) the majority of the fibers were within the 300–400 nm range with an array of very fine fibers around 100 nm connecting the thicker fibers. Higher concentrations of PCL yielded less fine fibers but overall a more even product with less variation in diameter. PLA fibers, on the other hand, were more uniform in average diameter and about half the size in diameter of PCL fibers from chloroform solutions of comparable concentrations.

Release of antibiotics into physiological buffer solution

Electrospun PCL and PLA fibers were loaded with one of two antibiotics very similar in structure (T and C, see Fig. 1 for their chemical structure) or one antibiotic/antifungal compound (AB) as model drugs

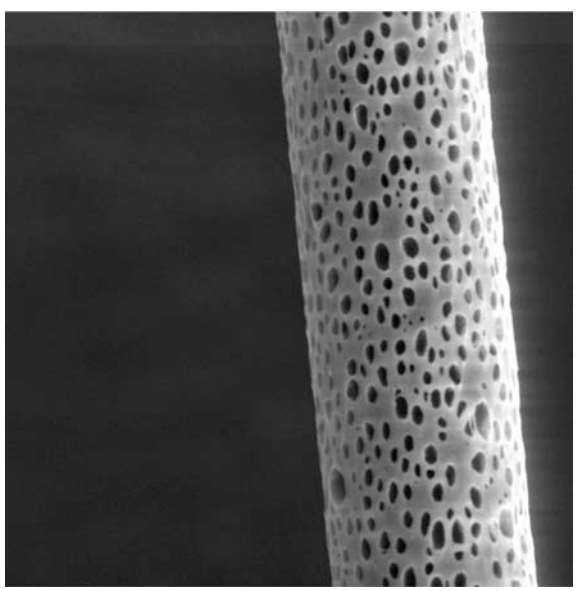


Fig. 3 Poly(ε-caprolactone) (PCL) fiber electrospun from 15 wt.% chloroform solution; magnification ×2,000 (diameter of fiber shown is approximately 6 μm)

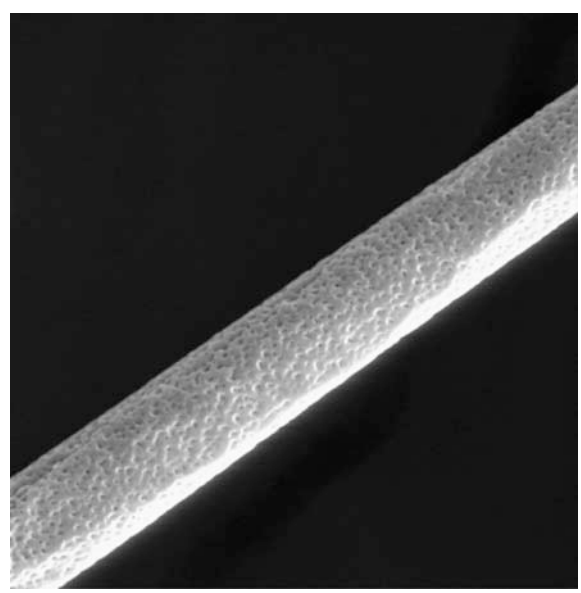


Fig. 4 Poly(L-lactic acid) (PLA) electrospun from 12 wt.% chloroform solution; magnification ×2,000 (diameter of fiber shown is approximately 3.2 μm)

for discharge into a physiologic buffer solution. All three compounds only very slowly or incompletely dissolved in the chloroform spinning solution. It was therefore necessary to use a small amount of a co-solvent (methanol). All fibers spun with methanol or DMF and from a slightly lower concentration of polymer (10 instead of 12–15 wt.%) were much smaller in diameter with less pronounced surface features. As an example, Fig. 5 shows a scanning electron micrograph of a fiber obtained from 10 wt.% PCL containing C predissolved in methanol. The indentations on the surface seemed to be fairly shallow and less distinct than those of PCL fibers from 15% chloroform alone (see Fig. 3).

Tetracycline drugs have a broad antibiotic spectrum that includes Gram-negative and Gram-positive bacteria. Amphotericin B is effective against fungi and yeast. The graph in Fig. 6 shows the release properties of PCL in comparison to PLA as well as the impact of chemical structure and properties of the drug towards its discharge characteristics. PCL released 75% of T within the first 20 min. While C is very similar in structure to T only approximately 30% of C was released within the same time. The discharge of C, although slow, continued over longer time than the release of T. It is possible that the effect is due to solubility differences of T and C in both the polymer solution as well as the buffer solution. PLA, on the other hand, discharged less than 10% total of

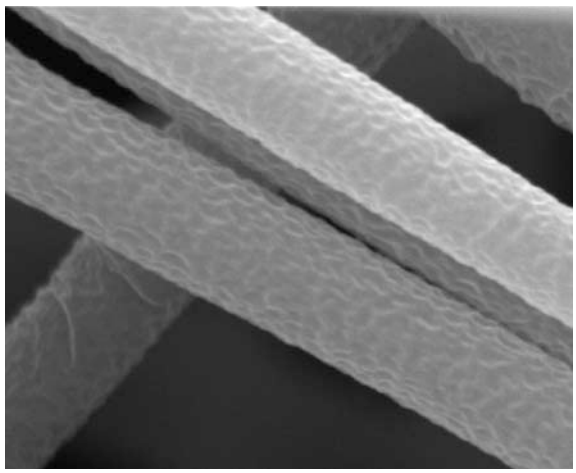


Fig. 5 Poly(ε-caprolactone) (PCL) fiber electrospun from 10% chloroform solution, containing chlorotetracycline hydrochloride (C) ×10,000 (average fiber diameter approximately 2.5 μm)

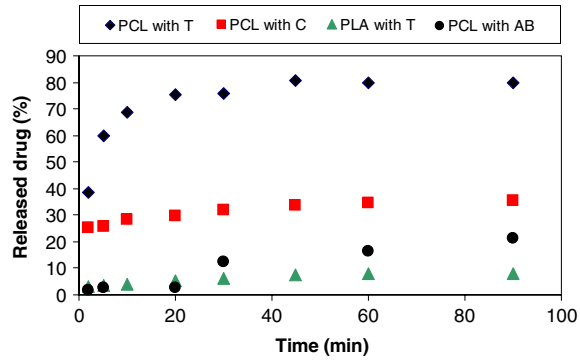


Fig. 6 Release of model drugs tetracycline hydrochloride (T), chlorotetracycline hydrochloride (C) and amphotericin B (AB) from poly(ε-caprolactone) (PCL) and T from poly(L-lactic acid (PLA) into a buffer solution of pH 7.35. Maximum standard deviation: ±4.0

the drug T, even over an extended period of time (data not shown). In this case, T seemed to be more readily available from PCL than from PLA, although the overall surface area of the PLA fibers is higher due to finer fiber diameters (see SEM in Figs. 3 and 4). The PCL surface appears to be more open than PLA which might play a role. Differences in the interaction of the drugs with the polymers are also important to take into consideration. In the case of AB a lower, slowly increasing quantity was released from PCL within 90 min (Fig. 6). Only after 3 h approximately 30% AB on the average was discharged into the buffer and the release slowed without reaching a clear plateau even after 24 h (data not shown). AB is larger in structure than T and C and only disperses in the buffer solution. Additionally, the co-solvent for spinning was DMF instead of methanol. It is very likely that all these factors impact the location and availability of the drug inside the fiber and its release rate.

Antimicrobial tests were performed using *Escherichia coli* and *Staphylococcus aureus* as a representative each of Gram-negative and Gram-positive bacteria, respectively. Results for *E. coli* are shown in Table 3 for T and C discharged from PCL fibers and compared to the unloaded control fiber. Both antibiotics showed similar effectiveness with 77% and 69% bacteria reduction after 10 min, for T and C respectively, and approximately 85–86% reduction after 1 h which is 3–4 times more than the control fiber without drugs. Both T and C were

Table 3 Effectiveness of T and C enclosed in PCL towards *E. coli*

Sample	Contact time (min)	Bacterial reduction (%)
PCL control	10	20.51
	30	24.00
	60	33.61
PCL + T	10	77.29
	30	84.28
	60	86.90
PCL + C	10	69.43
	30	81.66
	60	85.15

more effective in regard to *S. aureus*. In this case approximately 95% bacterial reduction was observed after only 10 min.

Formation of PLA–PCL bicomponent fibers and release of antibiotics

Figure 7 shows a SEM picture of an electrospun bicomponent fiber from PCL–PLA (1:1) from 12 wt.% chloroform solution. The porous surface of the fiber resembled closer the PLA single-component fibers with areas of somewhat more irregular and seemingly deeper indentations. Fiber diameters were rather large and varied from 1 to 4 μm with only very few fibers in the nanometer range.

In Fig. 8 a group of fibers spun from 12 wt.% PCL–PLA (1:3) is depicted. Interestingly in this case the fiber surface appeared to be more influenced by the PCL portion of the spinning solution although the composition would suggest otherwise. Under the applied conditions an array of thicker fibers were formed with diameters of approximately 1.5–3 μm (as shown in Fig. 8) and finer fibers with diameters ranging from 300 to 500 nm. Both types of fibers appeared uniform along the fiber axis and did not differ visually in their surface characteristics.

Delivery of model therapeutic compounds was studied from bicomponent PCL–PLA fibers of composition 3:1; 1:1; and 1:3, respectively. The results are graphically presented in Fig. 9. It was observed that fibers solely made from PLA and bicomponent fibers containing a higher portion of PLA than PCL discharged little of the drug, reaching a plateau at approximately 10% or lower after 30 min (PCL–PLA, 1:3), thus behaved similar to fiber mats containing only PLA. The micrographic pictures of these materials showed small indentations of fairly even size that might not reach deeply into the fiber interior. Fibers made from PCL alone or in bicomponent form with

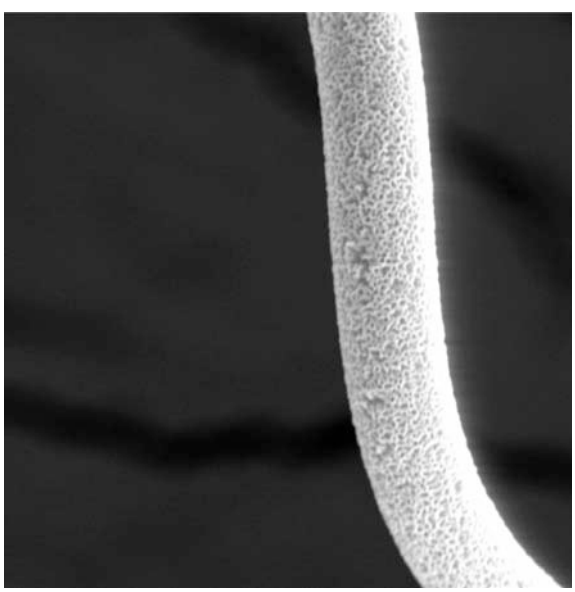


Fig. 7 About 12 wt.% poly(ε-caprolactone)-poly(L-lactic acid) co-electrospun fiber (1:1); magnification ×2,000 (diameter of fiber shown is approximately 3 μm)

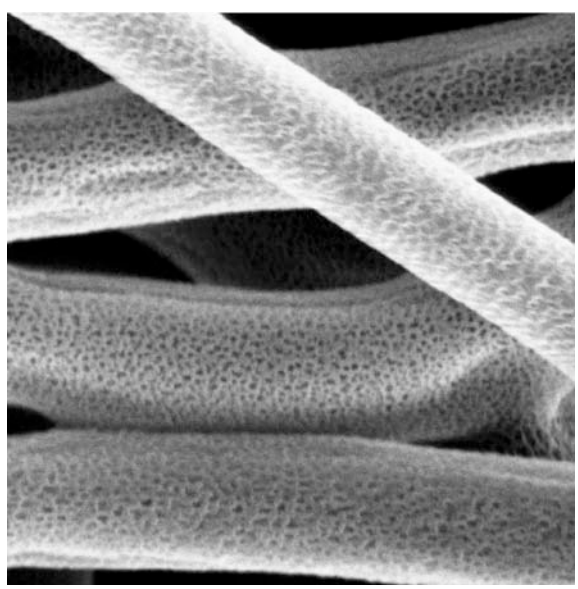


Fig. 8 About 12 wt.% poly(ε-caprolactone)-poly(L-lactic acid) (1:3) co-electrospun; magnification ×5,000 (diameter of fiber shown is approximately 2 μm)

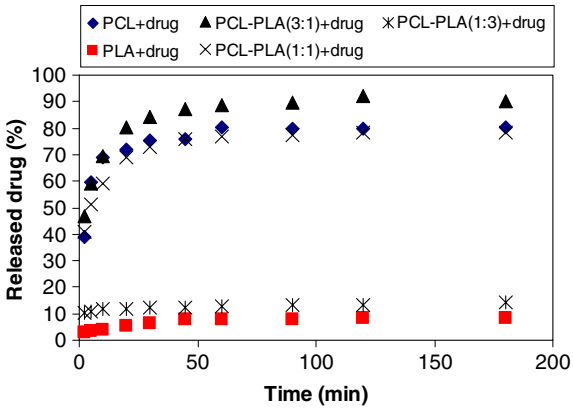


Fig. 9 Time-dependent delivery of tetracycline hydrochloride (T) from electrospun fiber mats into tris-buffer of pH 7.35. Maximum standard deviation: ±2.0

equal or dominant PCL ratio, on the other hand, rapidly liberated more than 70% of the drug within the first 30 min before reaching a plateau value. Such fast release would probably be less useful for medical applications. The serrations of these fibers appeared larger and of less even shape (compare to Fig. 3). In future research the gaps between polymer ratios of PCL:PLA will be narrowed because it seems possible to target a specific range of suitable release rates for a therapeutic compound based on solution composition. It is clearly helpful to first establish a fiber/drug system individually and subsequently build on the results. Varying the amount or type of the co-solvent also promises to be a regulator for a specific system (unpublished data).

Conclusions

PCL and PLA fibers as well as bicomponent PCL–PLA fibers were electrospun at concentrations in the range of 9–15 wt.% and loaded with three different antibiotics. Chloroform proved to be the most suitable solvent for this purpose. Fiber properties were established by determining the viscosity, shear behavior, surface tension and the solvent evaporation behavior of the spinning solution. All fibers showed variations in surface morphology. Depending on the spinning system indentations of various sizes, depths and shapes were observed. The most uniform bicomponent PCL–PLA fibers with serrated surfaces and the least variation in diameter were obtained from 12 wt.% chloroform solutions at ratios of 1:1, 3:1 or 1:3. The fibers were exposed to a buffer solution of pH 7.35 to study the discharge rate of the incorporated antibiotics. PCL released the drugs fairly fast and nearly completely, while PLA could hold the drugs much longer. Bicomponent fibers of PCL–PLA behaved in a similar manner as the dominant polymer in the bicomponent mat with release characteristics falling in-between pure PCL and pure PLA fibers. It seems very likely that their release behavior can be shaped by careful fiber design in which the effect of each component in the spinning system is taken into account.

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References

Bognitzki M, Czado W, Frese T, Schaper A, Hellwig M, Steinhardt M, Greiner A, Wendorff JH (2001) Nanostructured fibers via electrospinning. Adv Mater 13(1): 70–72

Boland ED, Wnek GE, Simpson DG, Pawlowski KJ, Bowlin GL (2001) Tailoring tissue engineering scaffolds by employing electrostatic processing techniques: a study of poly(glycolic acid). J Macromol Sci 38:1231–1243

Buschle-Diller G, Erdonmez F (2003) Electrospinning of nanofibers from biodegradable polymers, Intern. Proceedings of Nonwovens Techn. Conf., Baltimore

Buschle-Diller G, Cooper J, Richardson T, Hawkins A (2005) Electrospun nanofibers from renewable resources. 229th ACS National Meeting, San Diego

Buschle-Diller G, Hawkins A, Cooper J (2006) Electrospun nanofibers from biopolymers and their biomedical applications. In: Edwards V, Buschle-Diller G, Goheen S (eds) Fiber modifications for specialty and medical applications. Springer, London, pp 67–80

Chu CC, von Fraunhofer JA, Greisler HP (1997) Wound closure materials and devices. CRC Press LLC, Boca Raton

Chung HY, Hall JRB, Gogins MA, Crofoot DG, Weik TM (2002) Polymer, polymer microfiber, polymer nanofiber and applications including filter structures PCT Int. Appl. WO 2001-US24948 20010809

Deitzel JM, Kleinmeyer J, Hirvonen JK, Beck TNC (2001) Controlled deposition of electrospun poly(ethylene oxide) fibers. Polymer 42:8163–8170

Doshi J, Reneker DH (1995) Electrospinning process and applications of electrospun fibers. J Electrostatics 35(2/3):151–160

Fong H, Chun I, Reneker DH (1999) Beaded nanofibers formed during electrospinning. Polymer 40:4585–4592

Gibson P (2003) Production and characterization of patterned electrospun fibrous membranes. Proceedings of Fiber Soc. Meeting, pp 42–49

Gibson P, Schroeder-Gibson H, Rivin D (2001) Transport properties of porous membranes on electrospun nanofibers. Colloids Surf A Physicochem Eng Asp 187–188:469–681

Gupta P, Wilkes GL (2003) Some investigations on the fiber formation by utilizing a side-by-side bicomponent electrospinning approach. Polymer 44:6353–6359

Hawkins A, Buschle-Diller G (2004) Characterization of polymer solutions intended for electrospinning. Fiber Society Fall Meeting, Ithaca

Hawkins A, Wood J, Buschle-Diller G (2004) Advances in electrospinning of biopolymers, AATCC Intern. Convention & Exhibition, Greenville

Huang ZM, Zhang YZ, Kotaki M, Ramakrishna S (2003) A review on polymer nanofibers by electrospinning and their applications in nanocomposites. Composites Sci Technol 63(15):2223–2253

Jiang H, Hu Y, Li Y, Zhao P, Zhu K, Chen W (2005) A facile technique to prepare biodegradable coaxial electrospun nanofibers for controlled release of bioactive agents. J Control Release 108:237–243

Kahol PK, Pinto NJ (2004) An EPR investigation of electrospun polyaniline-polyethylene oxide blends. Synth Met 140:269–272

Kenawy E-R, Abdel-Hay FI, El-Newehy MH, Wnek GE (2007) Controlled release of ketoprofen from electrospun poly(vinyl alcohol) nanofibers. Mat Sci Eng A 459:390–396

Kewany ER, Bowlin GL, Mansfield K, Layman J, Simpson DG, Sanders EH, Wnek GE (2002) Release of tetracycline hydrochloride from electrospun poly(ethylene-co-vinyl acetate), poly(lactic acid) and a blend. J Control Release 81(1–2):57–64

Kidoaki S, Kwon IK, Mattsuda T (2004) Mesoscopic spatial design of nano- and microfiber meshes for tissue-engineering matrix and scaffold based on newly devised multilayering and mixing electrospinning techniques. Biomaterials 26:37–46

Kim K, Luu YK, Chang C, Fang D, Hsiao BS, Chu B, Hadjiargyrou M (2004) Incorporation and controlled release of a hydrophilic antibiotic using poly(lactide-co-glycolide)-based electrospun nanofibrous scaffolds. J Control Release 98(1):47–56

Kit K, Jagannathan S (2002) Electrospinning of nylons, poly(ethylene terephthalate) and their blends. Fiber Society Fall Technical Meeting, Book of Abstracts, 25

Lee SH, Yoon JW, Moon H (2002) Continuous nanofibers manufactured by electrospinning technique. Macromol Res 10(5):282–285

Li L, Hsieh YL (2005) Ultra-fine polyelectrolyte fibers from electrospinning of poly(acrylic acid). Polymer 46:5133–5139

Luong-Van E, Grøndahl L, Chua KN, Leong KW, Nurcombe V, Cool SM (2006) Controlled release of heparin from poly(ε-caprolactone) electrospun fibers. Biomaterials 27:2042–2050

Luu YK, Kim K, Hsioa BS, Chu B, Hadjiargyrou M (2003) Development of a nanostructured DNA delivery scaffold via electrospinning of PLGA and PLA-PEG block copolymers. J Control Release 89(2):341–353

Reneker DH, Chun I (1996) Nanometer diameter fibres of polymer, produced by electrospinning. Nanotechnol 7(3):216–223

Simpson DG, Bowlin GL, Wnek GE, Stevens PJ, Carr ME, Matthews JA, Rajendran S (2004) Electroprocessed collagen and tissue engineering. US Patent application 0040037813

Shin YM, Hohman MM, Brenner MP, Rutledge GC (2001) Experimental characterization of electrospinning: the electrically forced jet and instabilities. Polymer 42(25):9955–9967

Smith LA, Ma PX (2004) Nano-fibrous scaffolds for tissue engineering. Colloids Surf B Biointerf 39(3):125–131

Spasova M, Manolova N, Paneva D, Rashkov I (2004) Preparation of chitosan-containing nanofibers by electrospinning of chitosan/poly(ethylene oxide) blend solutions. e-Polymers, no. 056

Taepaiboon P, Rungsardthong U, Supaphol P (2006) Drug-loaded electrospun mats of poly(vinyl alcohol) fibres and their release characteristics of four model drugs. Nanotechnology 17:2317–2329

Taepaiboon P, Rungsardthong U, Supaphol P (2007) Vitamin-loaded electrospun cellulose acetate nanofiber mats as transdermal and dermal therapeutic agents of vitamin A acid and vitamin E. Eur J Pharma Biopharma 67:387–397

Theron SA, Zussman E, Yarin AL (2004) Experimental investigation of the governing parameters in the electrospinning of polymer solutions. Polymer 45:2017–2030

Tungprapa S, Jangchud I, Supaphol P (2007) Release characteristics of four model drugs from drug-loaded electrospun cellulose acetate fiber mats. Polymer 48:5030–5041

Yoshimoto H, Shin YM, Terai H, Vacanti JP (2003) A biodegradable nanofiber scaffold by electrospinning and its potential for bone tissue engineering. Biomaterials 24:2077–2082

Zeng J, Yang L, Liang Q, Zhang X, Guan H, Xu X, Chen X, Jing X (2005) Influence of the drug release compatibility with polymer solution on the release kinetics of electrospun fiber formation. J Control Release 105(1–2):43–51

Zong X, Kim K, Fang D, Ran S, Hsaio BS, Chu B (2002) Structure and process relationship of electrospun bioabsorbable nanofiber membranes. Polymer 43(16):4403–4412