

REVIEW ARTICLE

Nanoparticulates with drug release based on temperature change

Nanočásticové systémy uvolňující léčivo při změně teploty

Miloslava Rabišková • Eva Koziolová • Johana Jirásková

Received 26 November 2014 / Accepted 21 December 2014

Summary

The stimuli-induced release systems are able to respond to an external stimulus resulting in drug release in a controlled pattern. The origin of the external stimuli may be of physical, chemical or biological nature. Thermo-responsive delivery systems respond to the change in temperature and they were mainly designed in order to be used in the cancer treatment method using elevated temperature, i.e. hyperthermia. The thermo-responsive systems can be divided into several groups, such as thermo-responsive hydrogel polymer systems, liposomes, nano- or microparticles, and polypeptide-drug conjugates. While liposomes are temperature-sensitive by their nature, the other systems are usually based on thermo-sensitive polymers, namely poly-(*N*-isopropyl-acrylamide). This article summarizes recently available items of information regarding thermo-responsive drug delivery.

Keywords: drug delivery system • thermo-responsive system • poly-(*N*-isopropyl-acrylamide) • liposome • nanoparticle • polypeptide-drug conjugate

Souhrn

Lékové transportní systémy reagující na vnější podněty jsou schopné uvolnit léčivou látku požadovaným řízeným způsobem v závislosti na spouštěcím mechanismu. Spouštěcí mechanismus může být fy-

zikální, chemické nebo biologické povahy. Termoresponzivní lékové transportní systémy odpovídají na změnu teploty a byly navrženy zejména k léčbě rakoviny metodou využívající působení zvýšené teploty, tj. hypertermii. Termoresponzivní systémy lze rozdělit do několika skupin, např. termoresponzivní hydrogelové polymerní systémy, lipozomy, nano- nebo mikročástice a polypeptidové konjugáty s léčivem. Zatímco lipozomy jsou citlivé na zvýšení teploty již svým složením, ostatní systémy jsou obvykle založené na termosenzitivních polymerech, zejména poly-(*N*-izopropyl-akrylamidu). Tento článek shrnuje poslední dostupné informace týkající se cíleného uvolňování léčiv v závislosti na změně teploty.

Klíčová slova: lékový transportní systém • termoresponzivní systém • poly-(*N*-izopropyl-akrylamid) • lipozom • nanočástice • peptidový konjugát s léčivem

Introduction

Controlled-release drug delivery systems (second and third generation dosage forms), which predetermine the drug release profiles and the drug bio-distribution due to the unique properties of the delivery system, have gained great interest in pharmaceutical research and development throughout the world. The drug release from the second generation drug delivery systems may have the profiles of a prolonged release, a delayed release or a pulsatile release. The third generation of the drug delivery systems may influence the drug bio-distribution and target the drug to the specific site (organ, tissue, cells).

Second generation dosage forms have been developed, introduced on the market and described in many articles^{1–5)}. They already contribute to safe and efficient pharmacotherapy. However, the third generation dosage forms based on a specific drug release in the demanded part of the body are the objective of present international research. These sophisticated systems can respond –

prof. PharmDr. Miloslava Rabišková, CSc. (✉) • J. Jirásková
Department of Pharmaceutical Technology
Faculty of Pharmacy, Charles University in Prague
Heyrovského 1203, 500 05 Hradec Králové
e-mail: rabiskom@faf.cuni.cz

E. Koziolová
Institute of Macromolecular Chemistry
Czech Academy of Science, Praha

i.e. release the included drug – to some stimuli. That is why these systems are also called stimuli-responsive systems⁶⁾.

The stimuli-induced delayed release systems, also known as “smart” or “intelligent” systems, are able to receive and process an external stimulus, to which they consequently respond, resulting in drug release (Fig. 1). The system response to the stimuli may be based on a change in the system properties such as the volume, the secondary structure, the phase, the solubility, or the permeation. The alteration of the system physical or chemical behaviour leads to the release of the incorporated drug in a controlled pattern. The origin of the external stimuli may be physical, e.g. temperature change, electric or magnetic field, ultrasound; chemical (pH or glucose change), or biological triggered by antigen or enzyme^{7, 8)}.

Temperature-induced release systems

The well-known fact that physiological temperature changes in the presence of pathogens or pyrogens was the main idea of developing the thermo-responsive (TR) systems. Another motive for designing TR systems is the cancer treatment method using the externally applied, induced hyperthermia targeted to the tumor tissue. The adjuvant effect of hyperthermia (heating the tumor tissues to temperatures by 2–8 °C higher than the body temperature) has been clinically proved in the treatment of solid tumors when it is combined with chemotherapy or radiotherapy. It is explained by the fact that the temperatures above 41–42 °C have a direct cytotoxic effect. In addition, hyperthermia usually increases the perfusion of most tumors, which may result in an enhancement of the drug delivery to the tumors^{10–12)}. The externally applied, induced hyperthermia may also serve as the stimuli to trigger the drug release from the thermo-sensitive systems. The ideal TR system stays intact at the physiological body temperature ($T_B = 37^\circ\text{C}$), the drug from the system is released after exceeding a certain threshold temperature of the system due to the induced hyperthermia – such a phenomenon is called the active targeting. This implies that after the administration of the drug-loaded TR system to the blood circulation, the drug is released only in the externally heated tumors. This method of the active drug targeting may fundamentally reduce the cytotoxic side-effects of anticancer medicaments and can enhance the efficiency of the tumor treatment^{13, 14)}.

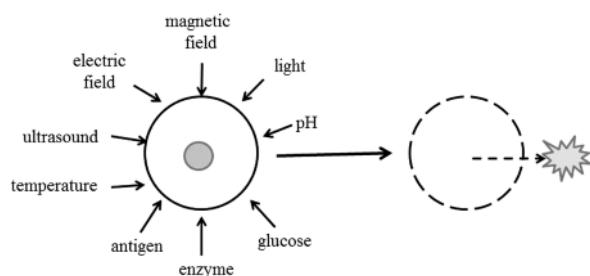


Fig. 1. The illustration of the stimuli-induced release from a drug-loaded particle (modified from⁹⁾)

Furthermore, the systems used in the anti-cancer therapy employ passive targeting to the tumors. The passive targeting is dependent on the properties of the drug delivery system and the pathologic conditions of the tumor neovasculature which is disorganized with dead ends, loops and openings into the perivascular space. The presence of the openings enhances the macromolecular or nanoparticulate permeation of the tumor. In addition, the absence of the tumor lymphatic drainage reduces the drug clearance. The occurrence of these two processes is generally known as the enhanced permeability and retention (EPR) effect^{15, 16)} as shown in Figure 2.

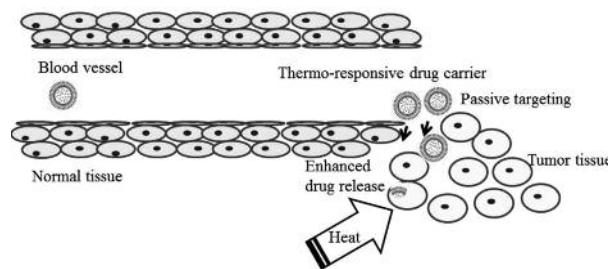


Fig. 2. The scheme of the passive targeting using the enhanced permeability and retention (EPR) effect and the drug release from the thermo-responsive drug carriers in the heated tumor (modified from²¹⁾)

However, passive drug targeting itself has failed to achieve increased clinical efficacy in humans when compared with the free drug as a result of several shortcomings. Accumulation of nanoparticles depends on the specific structure of tumor vasculature¹⁷⁾, it competes with uptake in the liver and spleen and less than 10% of the injected dose accumulates in the tumor¹⁸⁾.

To solve this problem, an additional approach for temperature-triggered drug delivery including a pre-hyperthermia treatment was proposed. This pre-hyperthermia method opens up the tumor vasculature for passive accumulation¹⁷⁾ and is followed by a second heat trigger for interstitial drug release¹⁹⁾. Such formulations might be further surface-modified for active targeting to the tumor vasculature or tumor cells²⁰⁾.

The thermo-responsive systems can be divided into several groups, such as TR hydrogel polymer systems, TR liposomes and other nano- or microparticles, and TR polypeptide-drug conjugates.

The thermo-sensitive hydrogel polymers are capable of reversible swelling/deswelling dependent on temperature changes. The solubility of these polymers has a non-linear relationship with temperature. Each thermo-responsive polymer is characterized by the specific sharp transition temperature, known as the lower critical solution temperature (LCST), at which the polymer solubility shifts from soluble to insoluble. Various synthetic and natural polymers show the thermo-responsive behaviour, such as synthetic poly(*N*-isopropyl-acrylamide) (pNIPAAm), polyethylene oxide-*co*-polypropylene oxide-*co*-polyethylene oxide, or semi-natural chitosan²²⁾.

The commonly used TR polymer is cross-linked

(pNIPAAm) (Fig. 3) due to the LCST (32°C) in an aqueous solution near the body temperature. Below the LCST the pNIPAAm segments are hydrated forming expanded structures enabling water to diffuse, the hydrogel absorbs water and swells. Above the LCST polymer chains dehydrate, water is squeezed out and the polymer shrinks, as it is shown in Figure 4. This swelling/deswelling phase transition leading to a volume change is caused by the disruption of the hydrogen bindings between the water molecules and the polymer due to the temperature increase and the consequent interaction within the polymer matrix resulting in the formation of the intra- and inter-molecular hydrogen bonding within the polymer chains. The addition of either hydrophilic or hydrophobic compounds may alter the value of the LCST of the mixture with the polymer²³⁾.

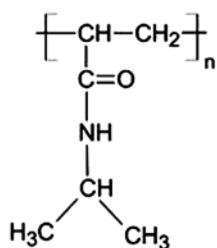


Fig. 3. The chemical structure of pNIPAAm

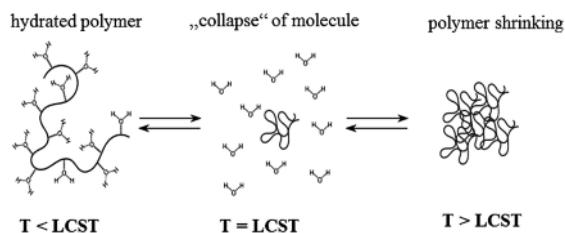


Fig. 4. The thermo-sensitive behaviour of the TR polymer pNIPAAm (modified from²⁴⁾)

TR ability of pNIPAAm has been applied in numerous temperature-induced drug delivery systems. The transformation process is reversible and repeatable, which enables the formation of the device with an “on-off” model of the drug release. The release mechanism lies in squeezing out the drug during the deswelling phase, or in entrapping the drug within the polymer matrix in the shrunken state, which is followed by amplified drug permeation during the swollen phase²⁵⁾. TR drug release profile depends on the system construction (matrix systems, micro- or nanoparticle systems), as it is described in Figure 5.

Hydrogel systems

Numerous hydrogel systems based on thermo-responsive polymers, especially pNIPAAm, have been synthesized. Thermo-responsive polymeric hydrogel may form polymeric matrices, membranes, or it may be incorporated into micro- or nanoparticle systems. The

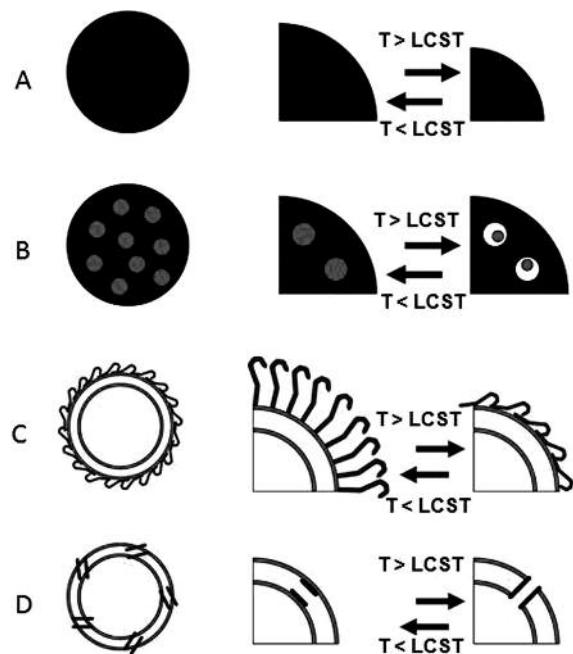


Fig. 5. The TR behaviour of various systems including the TR polymer: A) The hydrogel matrices squeeze out the drug during the hydrogel shrinking above the LCST²⁶⁾; B) the grafts of TR polymers within microspheres form pores above the LCST resulting in the drug release²⁵⁾; C) the micro- or nano-particles with the surface modified with TR polymer chains²⁷⁾, below the LCST polymer chains in a form of open coils allow drug permeation, above the LCST the collapsed polymer chains cover the surface and reduce drug permeability; D) the microcapsules with TR gates in the membrane release the drug above the LCST²⁸⁻³⁰⁾.

pNIPAAm hydrogel with the LCST = 32°C has been further modified in order to obtain a system with optimal characteristics such as an accelerated response to temperature^{27, 28)} or an increased transition temperature of the system³³⁾ in order to employ the induced hyperthermia in cancer-treatment.

Okuyama et al. presented the thermo-responsive indomethacin release from the cross-linked poly(*N*-isopropylacrylamide-*co*-alkyl methacrylate) polymer. The polymer formed a hydrogel matrix with the dispersed drug, which swells below the LCST and the drug is released just via diffusion. Upon a temperature increase, the polymer becomes hydrophobic and shrinks. The shrinking induces the burst drug release due to the “squeezing effect” – the flux of water with the dissolved/dispersed drug is squeezed out from the polymer matrix³³⁾.

The methods that accelerate structural changes of polymer in response to temperature changes have been reported by Kaneko and some other authors^{31, 34, 35)}. Grafted the cross-linked pNIPAAm hydrogels by the free mobile linear pNIPAAm chains allows a rapid swelling/deswelling transition of the system. The swelling/deswelling kinetics of the grafted gels is dependent on the graft chain lengths, as well as the drug release profiles of the drugs with large molecular weights.

Another method of designing TR drug delivery systems was reported by Oh et al. They used a mixture of

polyethylene oxide-polypropylene oxide-polyethylene oxide triblock copolymer and polyvinyl alcohol, which forms a complex gel via hydrogen bonding in an aqueous solution. These intra/intermolecular interactions are thermo-sensitive. The gel beads of the polymer were stabilized by coating with the poly-(lactide-*co*-glycolide) membrane. The polymer ratio determines the swelling transition properties of the gel. The model drug, acetoaminophen, was released at the temperatures between 35 and 40 °C cit.³⁵.

The incorporation of TR polymers into the micro- or nano-particulate systems presents another possibility of designing a TR system with suitable properties.

Thermo-responsive liposomes

Liposomes are spherical nanoscale vehicles formed by a membrane bilayer usually composed of phospholipids. The membrane encloses an aqueous core that can be used to encapsulate hydrophilic drugs, whereas lipophilic drugs can be incorporated into the membrane. The liposome phospholipid membranes undergo phase transitions at the particular temperature (transition temperature T_T) from a solid gel phase to a liquid crystalline phase, which increases the hydration of the bilayered membrane leading to a potential drug leakage from the liposome^{36, 37}, as it is shown in Figure 6.

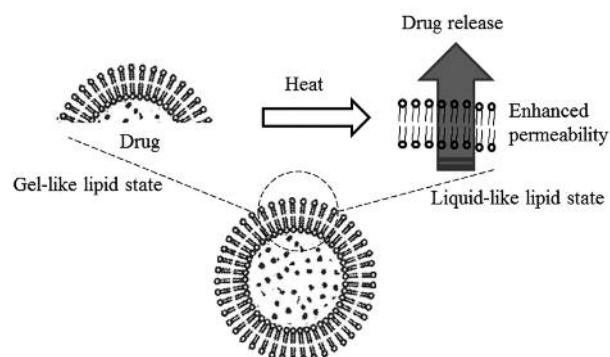


Fig. 6. The scheme of the drug release from thermo-responsive liposomes (modified from²¹)

The first TR system designed by Yatvin et al. was based on liposomes of 1,2-dipalmitoylphosphatidylcholine (DPPC). The gel-to-liquid crystalline phase transition of the phospholipid membrane occurred at the temperatures (41 °C) above the body temperature T_B . The liposomes became consequently leaky to small water-soluble molecules. Adding a small portion of the co-lipid distearoylphosphatidylcholine (DSPC) to the liposomal membrane helped to adjust the transition temperature to the desired value^{38, 39}.

At the beginning of the 21st century, Needham et al. prepared the TR liposomes with entrapped doxorubicin. They also used DPPC phospholipid as the structural material of the liposome membrane, but with the addition of 1-palmitoyl-2-hydroxy-*sn*-glycero-3-phosphocholine. Novel liposome formulation improved the delivery properties of the system. The transition temperature was lower (39–40 °C) compared to the

transition temperature of the pure DPPC liposome (42 °C); thus the release of the entrapped doxorubicin was exceptionally fast when the system was exposed to the temperature of 42 °C. The ultrafast drug release is regarded to be absolutely essential to enhance the drug permeation of the heated tumor due to the short circulating passage time in the tumor. *In vivo* testing in a human squamous cell carcinoma tumor xenograft model proved the complete response of all tumors and no tumor regrowth was reported during the following 60 days³⁹. The TR liposome formulation entered clinical trials, which was a breakthrough in the field visible by approximately 300 citations of the original paper⁴⁰.

The incorporation of lysolipids into the membrane bilayer of liposomes introduced a novel approach in the development of clinically usable TR liposome formulations. The surfactant lyso-PC mediates drug release by formation of lysolipid-stabilized membrane pores⁴¹. The release rate of doxorubicin from lysolipid containing TR liposomes at 41.3 °C was 80% in 20 seconds⁴². This type of liposomes with encapsulated doxorubicin Thermodox® designed by Celsion Corporation is in the Phase III clinical trials to treat hepatocellular carcinoma⁴².

In 2004, a new liposomal formulation composed of DPPC, DSPC and 1,2-dipalmitoyl-*sn*-glycero-3-phosphodiglycerol (DPPG₂) was reported by Lindner et al.⁴³. DPPG₂ is a synthetic phospholipid with a molecular weight close to the natural one, differing by one additional glycerol molecule in order to prolong the circulation half-life. The release rate of doxorubicin was very fast at the temperature of 42 °C. Later on, Lindner et al. used a soluble lysolipid to modify the liposome membrane. The phase transition of the designed liposomes occurred at 41–42 °C exhibiting a rapid and efficient drug release in response to a temperature increase, which was proved *in vivo* in rats and hamsters to treat amelanotic melanoma⁴⁴.

A sterically stabilized liposomal formulation modified with PEG was developed for improved stability in serum and a better *in vivo* half-life. This formulation had very good stability in serum at 37 °C and the maximum drug release occurred at 42 °C. A two-step treatment, i.e. pre-hyperthermia to open up the tumor vasculature and second hyperthermia after passive accumulation in tumor tissue allowed drug release for precise intratumoral drug delivery¹⁹.

Different approach to improve the characteristics of the drug delivery from the TR liposomes was presented by Tagami et al. TR liposomes with a simplified formulation were prepared by the incorporation of the non-ionic surfactant Brij78 composed of the PEGylated acyl chain (stearyl ether) to the liposome bilayer of DPPC – the molar ratio of DPPC:Brij78 was 96 : 4. The minority component of Brij78 is thought to prolong the liposome circulation and improve the system pharmacokinetics. The drug release from the liposome induced by a mild hyperthermia (40–42 °C) was rapid (100% release in 2–3 minutes). Liposomes had a prolonged circulation time and they were relatively stable in serum at 37 °C temperature. The *in vivo* efficiency and the minimal toxicity of the system were

proved using the mouse mammary carcinoma implanted to mice. Doxorubicin uptake into the cells of the heated tumor was 5.2-fold higher when compared to free doxorubicin⁴⁵⁾.

Temperature-induced response of liposomes may also be accomplished by incorporating the TR polymers (e.g. pNIPAAm) into the liposome structure. The fixation of the relatively hydrophilic TR polymers in the liposome membrane is provided via hydrophobic anchors (e.g. octadecyl/pyrene groups), which are incorporated to the polymer backbone. The position of the anchor within the backbone may influence the drug release behaviour. Below the LCST, the hydrophilic polymeric chains surround the liposome membrane improving the liposome stability. Above the LCST, the polymer dehydration and deswelling intensify the hydrophobic character of the liposome membrane leading to the exposure of the uncovered lipid membrane surface and the liposome destabilization and aggregation (Fig. 7). The TR transformation of the liposome surface properties permits the liposomes to interact with the cells. The TR polymers attached to the liposome membrane can also improve the liposome surface properties. Below the LCST, the highly hydrophilic polymer chains surrounding the liposome membrane suppress the interaction between the liposomes and proteins or cells resulting in an improved liposome stability *in vivo*⁴⁶⁾.

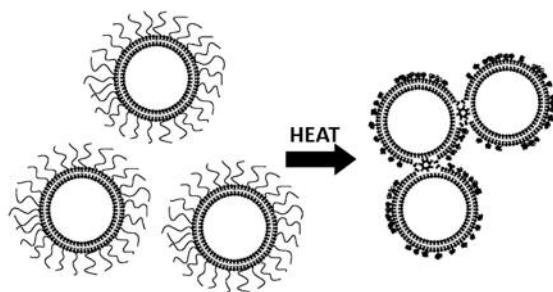


Fig. 7. The scheme of the thermo-responsive behaviour of the liposomes modified with the TR polymers⁴⁶⁾

Yoshino et al. prepared phosphatidylcholine liposomes modified with the TR polymers. The blocks of copolymers of NIPAAm and *N*-isopropylmethacrylamide (NIPMAM) were fixed in the liposome membrane using the anchors bound to the terminal end of the polymer chain in order to obtain the TR system, which would release the drug at the temperatures slightly higher than the body temperature enabling the targeted-specific treatment with the employment of mild hyperthermia. Above the LCST of the copolymer (40 °C), the dehydrated polymeric chains destabilized the liposome membrane leading to the liposome aggregation and the drug release²⁷⁾.

Other thermo-responsive micro- and nano-particles

Incorporating TR polymer material to the structure of micro- or nanoparticles opens the design opportunities of the drug delivery systems with targeted release. The

properties and the size of these systems enable the target-specific delivery of problematic drugs (cytotoxic anti-cancer drugs, macromolecules, insoluble compounds) and genes in order to improve the treatment efficiency and minimize the undesirable side-effects. Microspheres and microcapsules, and various types of nano-particles such as polymeric micelles or dendrimers may be utilized for the temperature-induced drug delivery¹⁴⁾.

The thermo-sensitive microspheres were designed by Curcio et al. The size of microspheres was 10–100 µm, they had a spherical shape and monolithic composition. They were prepared by grafting of NIPAAm with gelatine and parallel polymer cross-linking with *N,N*'-methylenebisacrylamide. Incorporation of hydrophilic gelatine units into the polymeric matrix of pNIPAAm leads to an increase in the LCST of the system (34.6–34.8 °C) compared to the LCST of the pure pNIPAAm (32 °C). Below the LCST the hydrogel swells allowing the enclosed model drug, diclofenac sodium salt, to diffuse slowly to the medium. The burst drug release occurs at 40 °C, when the temperature is higher than the LCST, the pNIPAAm segments deswell and the system collapses. To obtain the delayed release, the system requires further optimization to prevent the diffusion of the drug during the swollen state²⁵⁾.

Chu et al. presented different designing of the thermo-sensitive microparticles – the core-shell microcapsules with the gating shell. The shell is represented by the outer porous membrane with the gates which are capable of responding to temperature. To obtain the TR gates, the linear polymer of pNIPAAm is grafted to the porous surface of the polyamide microcapsules loaded with the model substance (NaCl or vitamin B₁₂) by the plasma-graft pore-filling polymerization method. At a low graft yield, a positive system response to the temperature was achieved, which means that the release rate increases with temperature (Fig. 8). The linear polymer chains of the gates become hydrophobic and coil up above the LCST allowing the encapsulated drug to diffuse out²⁸⁾.

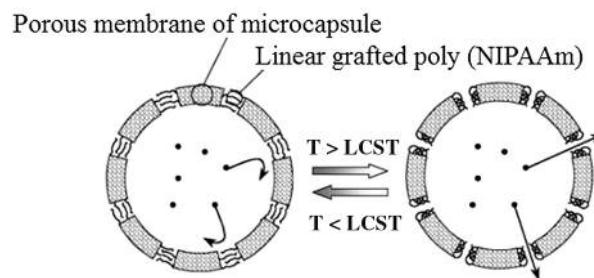


Fig. 8. The thermo-responsive release principle of the microcapsules with a porous membrane and the TR polymeric gates (modified from²⁸⁾)

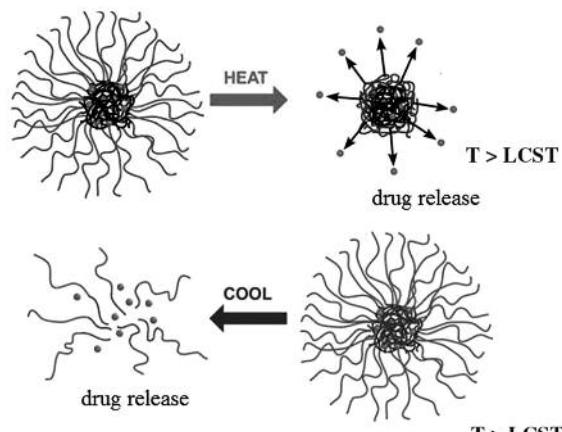
The thermo-sensitive polymeric micelles have recently been subjected to intensive research. The polymeric micelle consists of the amphiphilic block copolymers, surfactants. The surfactants contain hydrophobic and hydrophilic groups within the same molecule. When immersed in an aqueous solution, the interaction between water and the amphiphilic molecules

causes a self-assembly of micelles with a specific structure of a hydrophobic core and a hydrophilic corona. To achieve the micelle self-assembly, the concentration of the surfactants in the solution needs to exceed a certain value – the critical micelle concentration⁴⁷⁾. After incorporating the TR polymer to the amphiphilic block copolymers, the drug may be released in response to the temperature.

The properties and the *in vivo* behaviour of polymeric micelles allow them to act as a drug carrier of the drug delivery systems with targeted release. The exceptional structural properties of the polymer micelles and their nano-size hinder the micelles from being detected by the reticuloendothelial system. The micelles may be passively targeted to cancerous or inflamed tissues through the EPR effect^{9, 48, 49)}.

The thermo-sensitive polymeric micelles may be divided into two categories: polymeric micelles with TR outer coronas and polymeric micelles with TR inner cores. The drug release from each system is based on a different mechanism, as it is shown in Figure 9¹⁶⁾.

A – thermo-responsive corona



B – thermo-responsive core

Fig. 9. The drug release from the TR polymeric micelles: A) polymeric micelles with TR outer coronas; B) polymeric micelles with TR inner core (modified from¹⁶⁾)

Polymeric micelles with TR outer coronas are preferentially applied *in vivo*, because a mild hyperthermia is easily attainable. Below the transition temperature the pNIPAAm corona is hydrated and the block copolymers form the micellar structure in an aqueous solution. Above the LCST of pNIPAAm (32 °C) the corona becomes hydrophobic because of the polymer chains dehydration. Thus the micelles aggregate and precipitate leading to the drug release¹⁶⁾.

Polymeric micelles with TR inner cores are composed of block copolymers with hydrophilic segments (e.g. PEG) and TR polymer segments (pNIPAAm). Below the LCST the molecules of copolymers are dispersed in the aqueous solution due to the water-solubility of both copolymer components. Above the LCST the TR polymer segments are dehydrated and hydrophobic resulting in an association of polymer chains. The associated chains form a micelle structure with a core and a corona, as it is

displayed in Figure 9B. After the administration of the polymeric micelles to the circulation, the temperature of the certain tissue or organ is decreased in order to obtain the TR drug release with the targeted-specific pattern. The demand of the temperature decrease treatment, hypothermia, limits the application of the system. The benefits of these systems are a fast drug release in response to the temperature, the water-soluble polymer chains are dispersed in the solution and the drug does not need to diffuse from the micelle. Furthermore, the micelle preparation does not require the use of organic solvents and the drug incorporation is facile. The polymeric micelles with TR inner cores composed of polyethylene glycol and poly(*N*-isopropylacrylamide) were designed by Neradovic et al.⁵⁰⁾.

Chung et al. prepared TR polymer micelles with outer coronas from the block copolymers of poly (*N*-isopropylacrylamide-*co*-butylmethacrylate (pNIPAAm-pBMA). The hydrophobic core formed by pBMA segments serves as a reservoir of the lipophilic anti-cancer drug – adriamycin, whereas the hydrophilic pNIPAAm chains respond to temperature changes. The efficiency of the system was tested on endothelial cell culture media. The micelles exhibited no cytotoxicity below pNIPAAm's LCST, when the drug is not released. At the temperature of 37 °C, an enhanced interaction between the precipitated micelles and the cells was reported implying that the adriamycin-loaded thermo-responsive micelles possess higher cytotoxic effect than free adriamycin⁵¹⁾. The authors also studied the intracellular distribution of the doxorubicin released from the drug-loaded TR micelles within the human breast cancer cell culture under the conditions of mild hyperthermia (42 °C). In comparison with free doxorubicin, the employment of the drug loaded in the TR micelles enhances significantly the cellular uptake of doxorubicin when exposed to a high temperature (42 °C). The improved doxorubicin uptake by cells may be caused by the hydrophobic interaction of the aggregated micelles with the cellular membrane above the LCST. On the other hand, below LCST the hydrated micelle corona prevents from micelle adhesion to the hydrophobic cellular membrane. The promoted cellular uptake of the aggregated micelles leads to an enhanced intracellular drug release in the heated tumor cells, which maximises the anti-cancer treatment and reduces the toxic effect of the chemotherapy^{41, 52)}.

To be employed in the human body, the system required further adjustment of the value of the LCST and more biocompatible material than the used hydrophobic polymers. Kohori et al. managed to increase the system biocompatibility and the LCST of the system to the value right above the body temperature by utilizing the copolymer of poly-(NIPAAm-*co*-*N,N*-dimethylacrylamide) and biodegradable polylactic acid with the LCST of 40 °C to form TR micelle containing adriamycin. The drug molecules were incorporated into the micelle structure during their formation^{53, 54)}. Instead of polylactic acid, the polymer of poly- ϵ -caprolactone or the copolymer of poly(lactide-*co*- ϵ -caprolactone) may also be used^{55, 56)}. By means of an induced hyperthermia at 42.5 °C, the system may be used for *in vivo* targeting.

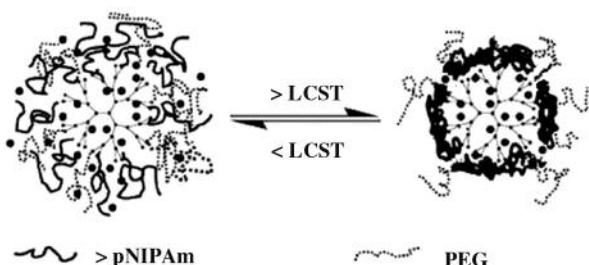


Fig. 10. The formation of the hydrophobic barrier on the surface of the TR dendrimer⁵⁷

Zhao et al. prepared PEGylated thermo-responsive dendrimers. Dendrimers are synthetic macromolecules, whose polymer chains emanate from a central core. The unique tree-like structure forms internal cavities that enable the encapsulation of water-soluble drugs. The polyamidoamine (PAMAM) polymer forming the dendrimer structure was polymerized with the thermo-responsive polymer of pNIPAAm and the copolymer was subsequently covalently linked with PEG in order to eliminate the dendrimer toxicity and immunogenicity and to prolong the circulation time. The encapsulated model drug, indomethacin, diffuses from the core below the LCST (30 °C), when the pNIPAAm chains are hydrated. Above the LCST (37 °C), the pNIPAAm chains become hydrophobic and shrink creating the water-impermeable barrier around the particle, which prevents the drug from diffusing, as it is shown in the Figure 10. The system response to the increase of the temperature is negative – the release rate decreases⁵⁷.

Thermo-responsive polypeptide-drug conjugates

TR characteristics of certain lipidic and polymeric materials are mentioned above. However, the thermo-responsive systems based on proteins have been designed as well. The biopolymers consisting of a repeated sequence of Val-Pro-Gly-Xaa-Gly (valine-proline-glycine-any amino acid except proline-glycine) pentapeptide, commonly known as elastin-like polypeptides (ELPs), undergo a phase transition at the specific temperature T_T . ELPs are water-soluble below their T_T . When the temperature within the material exceeds the T_T , the ELPs become hydrophobic leading to the protein collapse and aggregation. This effect was first described by Urry et al.⁵⁸.

The T_T of the ELPs is dependent on the ELPs sequence, the chain length, and the concentration. The equation of the three parameters which quantitatively predicts the value of T_T was reported. By varying these three parameters the desired T_T may be obtained⁵⁹.

By employing TR ELPs as a drug delivery vehicle, the targeted release to tumors may be achieved using externally applied hyperthermia. Furthermore, the ELPs are soluble polymers, which tend to accumulate in tumors due to the EPR effect. Raucher et al. conjugated the anti-cancer drug doxorubicin to the macromolecule of ELPs to obtain an anti-tumor drug system with targeted release. Their polypeptide-drug conjugate undergoes the phase transition at mild hyperthermic

temperatures (42 °C). The system is composed of four functional domains (Fig. 11). The derivative of doxorubicin is covalently attached to the enzymatically cleavable tetrapeptide, Gly-Phe-Leu-Gly (GPLG, glycine-phenylalanine, leucine, glycine). After the cellular uptake of the system, the GPLG, which is a substrate for the lysosomal protease, is degraded in the lysosomal environment allowing the drug to release. The GPLG is bound to the thermo-responsive ELPs macromolecule. The terminal part of ELPs is modified with the cell penetrating peptide Tat consisting of 11 amino acids which enhances the system binding to the plasmatic membrane. Upon the exposure to a mild hyperthermia (42 °C), the ELPs collapse and aggregate. The protein aggregation and the presence of the Tat significantly enhance the cell uptake of the system. The fact that the liberation of the cytotoxic doxorubicin is intracellular predetermines the system to be highly effective with reduced side-effects^{60, 61}. Furthermore, the authors reported that the Tat peptide fused with ELP prevents the ovarian cancer cells from adhering, spreading, invading and migrating *in vitro* and *in vivo*. It is known that the cancer cell migration and their attachment to the cells play the key role in tumor dissemination and metastasis. Therefore the interaction of the Tat peptide-ELP conjugate with the tumor cells significantly limits the tumor cell migration, which caused the 80% reduction of the overall tumor mass in the *in vivo* mouse model⁶².

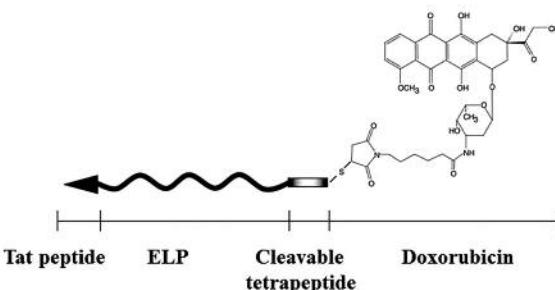


Fig. 11. The scheme of the ELP-based doxorubicin delivery vector (modified from⁶⁰)

In addition, the authors prepared the TR drug-ELP conjugate consisting of the TR ELP fused with a peptide drug and with the cell penetrating Bac peptide that helps the system to be delivered to the nucleus⁶³.

Meyer et al. designed a similar ELP-drug conjugate system based on the thermo-responsive ELPs and studied the system biodistribution *in vivo* using the human ovarian carcinoma implanted in mice. The rhodamine-labelled ELPs were applied to mice and the system accumulation in the tumor or the heated tumor was studied with the fluorescence videomicroscopy. The ELP-based system showed a 2-fold higher accumulation in the heated tumor than in the non-heated tumor tissue, which may be explained by the ELPs soluble-to-insoluble transition and their subsequent aggregation above the transition temperature leading to the ELP attachment to the cell membrane^{64, 65}. In comparison with the TR copolymer of p(NIPAAm-co-AAm) the

ELPs showed enhanced accumulation in the heated tumor cells^{66, 67}.

Conclusion

In conclusion, TR systems may represent a novel approach to drug targeting to a specific site, e.g. the tumor tissue. The drug release from TR dosage forms is induced and controlled via temperature variation. Up to now, numerous designs of TR systems have been reported. Certain systems were loaded with a drug and consequently subjected to the *in vitro* release testing. In order to prove the TR drug dosage form efficiency, the TR systems were added to cell culture models (e.g. of the tumor tissue). Furthermore, the biodistribution and *in vivo* system efficiency were tested using mice and other model animals. When compared with the free anti-cancer drug, the certain TR systems loaded with the same active substance showed a higher concentration in the tumor tissue and lower toxicity resulting in an increase in cancer therapy efficiency. Finally, the TR liposomes Thermodox® are currently undergoing the Phase III clinical trials for the treatment of liver cancer, so the complete clinical studies of the system are expected to be published soon.

Conflicts of interest: none.

References

1. Rabišková M., Fričová V. Perorální lékové formy s řízeným uvolňováním léčiv. Prakt. lékáren. 2008; 4, 212–216.
2. Dostálová M., Rabišková M. Mukoadhezivní orální tablety – moderní léková forma s řízeným uvolňováním léčiva. Čes. slov. Farm. 2000; 49, 55–61.
3. Rabišková M. Moderní lékové formy pro orální a perorální aplikaci. Bratislava: Farmaceutická fakulta Univerzity Komenského 2009.
4. Dvořáčková K., Rabišková M. Vaginální aplikace léčiv – nové směry. Praktické lékárenství 2006; 2, 93–97.
5. Bautzová T., Rabišková M., Lamprecht A. Multiparticulate systems containing 5-ASA for the treatment of inflammatory bowel disease. Drug Dev. Ind. Pharm. 2011; 37, 1100–1109.
6. Rabišková M. Nanočástice pro lékové formy. Remedie 2007; 17, 495–501.
7. Roy D., Cambre J. N., Sumerlin B. S. Future perspectives and recent advances in stimuli-responsive materials. Prog. Polym. Sci. 2010; 35, 278–301.
8. Gupta P., Verman K., Garg S. Hydrogels: from controlled release to pH-responsive drug delivery. Drug Discov. Today 2002; 7, 569–579.
9. Fleige E., Quadir M. A., Haag R. Stimuli-responsive polymeric nanocarriers for the controlled transport of active compounds: Concepts and applications. Adv. Drug Deliv. Rev. 2012; 64, 866–884.
10. Engin K. Biological rationale and clinical experience with hyperthermia. Control. Clin. Trials 1996; 17, 316–342.
11. Wust P., Hildebrand B., Sreenivasa G. Hyperthermia in combined treatment of cancer. Lancet Oncology 2002; 3, 487–497.
12. Gaber M. H., Wu N. Z., Hong K. L. Thermosensitive liposomes: Extravasation and release of contents in tumor microvascular networks. Int. J. Radiat. Oncol. 1996; 36, 1177–1187.
13. Meyer D. E., Shin B. C., Kong H. A. Drug targeting using thermally responsive polymers and local hyperthermia. J. Control. Rel. 2001; 74, 213–224.
14. Ganta S., Devalapally H., Shahiwala A. A review of stimuli-responsive nanocarriers for drug and gene delivery. J. Control. Rel. 2008; 126, 187–204.
15. Maeda H., Wu J., Sawa T. Tumor vascular permeability and the EPR effect in macromolecular therapeutics: a review. J. Control. Rel. 2000; 65, 271–284.
16. Nakayama M., Okano T. Multi-targeting cancer chemotherapy using temperature-responsive drug carrier systems. React. Funct. Polym. 2011; 71, 235–244.
17. Li L., Ten Hagen T. L., Bolkestein M. Improved intratumoral nanoparticle extravasation and penetration by mild hyperthermia. J. Control. Rel. 2013; 167, 130–137.
18. Harrington K. J., Mohammadtaghi S., Uster P. S. Effective targeting of solid tumors in patients with locally advanced cancer by radiolabeled pegylated liposomes. Clin. Cancer Res. 2001; 7, 243–254.
19. Li L., Ten Hagen T. L., Haeri A. A novel two-step mild hyperthermia for advanced liposomal chemotherapy. J. Control. Rel. 2014; 174, 202–208.
20. Dicheva B. M., Koning G. A. Targeted thermosensitive liposomes: an attractive novel approach for increased drug delivery of solid tumors. Expert Opin. Drug Deliv. 2014; 11, 83–100.
21. Marsh D. General features of phospholipid phase transitions. Chemistry and Physics of Lipids 1991; 57, 109–120.
22. Klouda L., Mikos A. G. Thermoresponsive hydrogels in biomedical applications. Eur. J. Pharm. Biopharm. 2008; 68, 34–45.
23. Bromberg L. E., Ron E. S. Temperature-responsive gels and thermogelling polymer matrices for protein and peptide delivery. Adv. Drug Deliv. Rev. 1998; 31, 197–221.
24. Abulataefeh S. R., Samer R., Spain A. G. Thermoresponsive polymer colloids for drug delivery and cancer therapy. Macromol. Biosci. 2011; 11, 1722–1734.
25. Curcio M., Spizzimi U. G., Iemma F. Grafted thermo-responsive gelatin microspheres as delivery systems in triggered drug release. Eur. J. Pharm. Biopharm. 2010; 76, 48–55.
26. Yoshida R., Sakai K., Okano T. Drug release profiles in the shrinking process of thermoresponsive poly(*N*-isopropylacrylamide-*co*-alkyl methacrylate) gels. Ind. Eng. Chem. Res. 1992; 31, 2339–2345.
27. Yoshino K., Kadowaki A., Takagishi T. Temperature sensitization of liposomes by use of *N*-isopropylacrylamide copolymers with varying transition endotherms. Bioconjugate Chem. 2004; 15, 1102–1109.
28. Chu L. Y., Park S. H., Yamaguchi T. Preparation of thermo-responsive core-shell microcapsules with a porous membrane and poly(*N*-isopropylacrylamide) gates. J. Membrane Sci. 2001; 192, 27–39.
29. Rabišková M. Využití nanočisticových systémů v medicíně. Remedie 2008; 18, 89–97.
30. Yoshida R., Uchida K., Kaneko Y. Comb-type grafted hydrogels with rapid de-swelling response to temperature-changes. Nature 1995; 374, 240–242.
31. Kaneko Y., Nakamura S., Sakai K. Rapid deswelling response of poly(*N*-isopropylacrylamide) hydrogels by the formation of water release channels using poly(ethylene oxide) graft chains. Macromolecules 1998; 31, 6099–6105.
32. Bae Y. H., Okano T., Kim, S. W. On/off thermocontrol of solute transport. 1. Temperature-dependence of swelling of *N*-isopropylacrylamide networks modified with hydrophobic components in water. Pharm. Res. 1991; 8, 531–537.
33. Okuyama Y., Yoshida R., Sakai K. Swelling controlled zero order and sigmoidal drug release from thermo-responsive poly(*N*-isopropylacrylamide-*co*-butyl methacrylate) hydrogel. J. Biomater. Sci. 1993; 4, 545–556.
34. Kaneko Y., Sakai K., Kikuchi A. Fast swelling/deswelling kinetics of comb-type grafted poly(*N*-isopropylacrylamide) hydrogels. Macromolecular Symposia 1996; 109, 41–53.
35. Oh K. S., Han S. K., Choi Y. W. Hydrogen-bonded polymer gel and its application as a temperature-sensitive drug delivery system. Biomaterials 2004; 25, 2393–2398.

36. Shaw A. W., Mclean M. A., Sligar S. G. Phospholipid phase transitions in homogeneous nanometer scale bilayer discs. *FEBS Letters* 2004; 556, 260–264.

37. Yatvin M., Weinstein J. N., Dennis W. H. Design of liposomes for enhanced local release of drugs by hyperthermia. *Science* 1978; 202, 1290–1293.

38. Weinstein J., Magin R. L., Yatvin M. Liposomes and local hyperthermia: selective delivery of methotrexate to heated tumors. *Science* 1979; 204, 188–191.

39. Needham D., Anyarambhatla G., Kong G. A new temperature-sensitive liposome for use with mild hyperthermia: Characterization and testing in a human tumor xenograft model. *Cancer Res.* 2000; 60, 1197–1201.

40. Kneidl B., Peller M., Winter G. Thermosensitive liposomal drug delivery systems: state of the art review. *Int. J. Nanomed.* 2014; 9, 4387–4398.

41. Landon C. D., Park J. Y., Needham D. Nanoscale drug delivery and hyperthermia: the materials design and preclinical and clinical testing of low temperature-sensitive liposomes used in combination with mild hyperthermia in the treatment of local cancer. *Open Nanomed. J.* 2011; 3, 38–64.

42. Phase 3 Study of ThermoDox with radiofrequency ablation (RFA) in treatment of hepatocellular carcinoma (HCC) [online]. CLINICALTRIALS.GOV, 2014-11-16 [cited 2014 11-16]. Available from: <http://www.clinicaltrials.gov/ct2/show/NCT00617981?term=ThermoDox&rank=3>.

43. Lindner L. H., Eichhorn M. E., Eibl H. Novel temperature-sensitive liposomes with prolonged circulation time. *Clin. Cancer Res.* 2004; 10, 2168–2178.

44. Li, L. Triggered content release from optimized stealth thermosensitive liposomes using mild hyperthermia. *J. Control. Rel.* 2010; 143, 274–279.

45. Tagami T., Ernsting M. J., Li S.-D. Efficient tumor regression by a single and low dose treatment with a novel and enhanced formulation of thermosensitive liposomal doxorubicin. *J. Control. Rel.* 2011; 152, 303–309.

46. Kono K. Thermosensitive polymer-modified liposomes. *Adv. Drug Deliv. Rev.* 2001; 53, 307–319.

47. Fletcher P. D. Self-assembly of micelles and microemulsions. *Cur. Opin. Colloid Interface Sci.* 1996; 1, 101–106.

48. Kwon G. S., Kataoka, K. Block-copolymer micelles as long-circulating drug vehicles. *Adv. Drug Deliv. Rev.* 1995; 16, 295–309.

49. Gaucher G., Dufresne M. H., Sant V. P. Block copolymer micelles: preparation, characterization and application in drug delivery. *J. Control. Rel.* 2005; 109, 169–188.

50. Neradovic D., Soga O., Van Nostrum C. S. The effect of the processing and formulation parameters on the size of nanoparticles based on block copolymers of poly(ethylene glycol) and poly(*N*-isopropylacrylamide) with and without hydrolytically sensitive groups. *Biomaterials* 2004; 25, 2409–2418.

51. Chung J. E., Yokoyama M., Okano T. Inner core segment design for drug delivery control of thermo-responsive polymeric micelles. *J. Control. Rel.* 2000; 65, 93–103.

52. Nakayama M., Chung J. E., Miyazaki T. Thermal modulation of intracellular drug distribution using thermoresponsive polymeric micelles. *React. Funct. Polym.* 2007; 67, 1398–1407.

53. Kohori F., Sakai K., Aoyagi T. Control of adriamycin cytotoxic activity using thermally responsive polymeric micelles composed of poly(*N*-isopropylacrylamide-co-N,N-dimethylacrylamide)-*b*-poly(d,l-lactide). *Colloids and Surfaces B: Biointerfaces* 1999; 16, 195–205.

54. Kohori F., Yokoyama M., Sakai K. Process design for efficient and controlled drug incorporation into polymeric micelle carrier systems. *J. Control. Rel.* 2002; 78, 155–163.

55. Nakayama M., Okano T., Miyazaki T. Molecular design of biodegradable polymeric micelles for temperature-responsive drug release. *J. Control. Rel.* 2006; 115, 46–56.

56. Qin S., Geng Y., Discher D. E. Temperature-controlled assembly and release from polymer vesicles of poly(ethylene oxide)-block-poly(*N*-isopropylacrylamide). *Adv. Mater.* 2006; 18, 2905–2909.

57. Zhao Y., Fan X., Liu D. PEGylated thermo-sensitive poly(amidoamine) dendritic drug delivery systems. *Int. J. Pharm.* 2011; 409, 229–236.

58. Urry D. W. Entropic elastic processes in protein mechanisms. I. Elastic structure due to an inverse temperature transition and elasticity due to internal chain dynamics. *J. Protein Chem.* 1988; 7, 1–34.

59. Meyer D. E., Chilkoti A. Quantification of the effects of chain length and concentration on the thermal behavior of elastin-like polypeptides. *Biomacromolecules* 2004; 5, 846–851.

60. Bidwell G. L., Davis A. N., Fokt I. A thermally targeted elastin-like polypeptide-doxorubicin conjugate overcomes drug resistance. *Invest. New Drug.* 2007; 25, 313–326.

61. Bidwell G. L., Raucher D. Application of thermally responsive polypeptides directed against c-Myc transcriptional function for cancer therapy. *Mol. Cancer Ther.* 2005; 4, 1076–1085.

62. Massodi I., Bidwell G. L., Davis A. N. Inhibition of ovarian cancer cell metastasis by a fusion polypeptide Tat-ELP. *Clin. Exper. Metastasis* 2009; 26, 251–260.

63. Massodi I., Moktan S., Rawat A. Inhibition of ovarian cancer cell proliferation by a cell cycle inhibitory peptide fused to a thermally responsive polypeptide carrier. *Int. J. Cancer* 2010; 126, 533–544.

64. Meyer D. E., Kong G. A., Dewhirst M. W. Targeting a genetically engineered elastin-like polypeptide to solid tumors by local hyperthermia. *Cancer Res.* 2001; 61, 1548–1554.

65. Chilkoti A., Dreher M. R., Meyer D. E. Design of thermally responsive, recombinant polypeptide carriers for targeted drug delivery. *Adv. Drug Deliv. Rev.* 2002; 54, 1093–1111.

66. Meyer D. E., Shin, S. C., Kong G. A. Drug targeting using thermally responsive polymers and local hyperthermia. *J. Control. Rel.* 2001; 74, 213–224.

67. Chilkoti A., Dreher M. R., Meyer D. E. Targeted drug delivery by thermally responsive polymers. *Adv. Drug Deliv. Rev.* 2002; 54, 613–630.