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Drug release from a N-vinylpyrrolidinone/acrylic acid lubricious hydrophilic coating

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Initial characterisation was performed on different crosslinked hydrogels to determine the suitability for coating applications. The test used in the initial stage of the work was swelling studies. From this test a suitable monomeric concentration was decided upon. The coating was applied to the Pebax® 3533 substrate using a dip coating/UV curing procedure. The coating was characterised by measuring the coating thickness, the relative viscosity, the density and the contact angle it made with the substrate. The coating was analysed using Ftir, optical microscopy and friction analysis to characterise the lubricious nature of the coating. From this analysis it was found that 2 coating cycles were necessary to give a good coverage thoughout the entire length of the substrate, and that the static coefficient was significantly reduced from that of the uncoated samples when the coating was hydrated for 10 min or more. Finally the release profile of aspirin was determined using UV spectroscopy, where it was found that the release rate could be controlled by varying the molecular weight of the crosslinking agents used.

1. Introduction

Biomaterials to be used for catheterisation in urinary, tracheal, and cardiovascular tracts, should have a surface that has good handling characteristics when dry, but which preferably becomes slippery upon contact with aqueous body fluids. Such a low-friction surface would enable easy insertion and removal from a patient. It would further prevent mechanical injury to the mucous membranes and would minimise discomfort to the patient [1].

Hydrophilic coatings are applied to medical devices to impart functional properties to the surface of a given device. Many medical devices are fabricated from conventional polymeric materials. Most polymeric materials exhibit poor wettability due to their low surface tensions. Even though some polymers exhibit relatively lubricious surfaces, for example polytetrafluoroethylene (PTFE), polyethylene (PE), and various polyamides (PAs), when in dynamic contact with the biological environment, they tend to irritate or damage surrounding tissue. By applying a hydrophilic coating, wettability or dynamic lubricity can be imparted to the existing device, thus minimising the trauma to the biological environment. Conroy [2] notes that the use of PTFE, glycerine or silicone fluid can be used to render the surface slippery. However these coating have the disadvantage that they can make the device difficult for the physician to manipulate, as silicone fluids and glycerine are greasy and sticky, and PTFE is slippery even when dry. It is for these reasons that hydrophilic coatings are finding increasing application in reducing an apparent surface friction of contact lenses and devices used in the field of minimally invasive surgery.

Minimising friction between a medical device and the surrounding tissue is a major design consideration. The extent to which hydrophilic coatings can reduce the coefficient of friction on the surface of a device is so dramatic; the term frictionless coating is commonly used to describe them [3]. A comparison of the static coefficient of friction for some commonly used materials is shown in Fig. 1.

Hydrogels are often used to improve or impart properties such as biocompatibility, wettability, and lubricity to systems deficient in these attributes. The extents to which the hydrogel can exhibit these properties are directly dependent on the amount of water that the hydrogel can imbibe into its molecular matrix, which is referred to as the degree of swelling. Several factors in the service environment have a direct bearing on the hydrogel's intended function and, may actually be used to control the response of the hydrogel in dynamic environments such as the human body. Typical factors that can induce such influence include temperature, pH,

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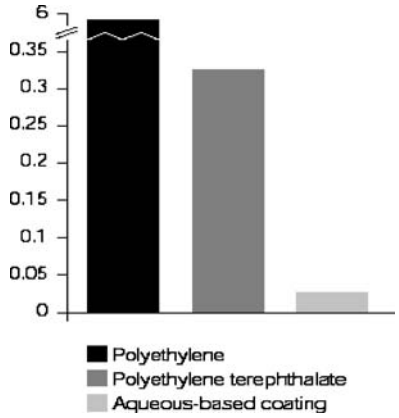

Figure 1 Comparision of co-efficient of friction of an aqueous-based coating, polyethylene and polyethylene terephthalate [2].

chemical agents, mechanical stress, electric potential, and magnetic fields [3].

Hydrogels are becoming increasingly important materials for pharmaceutical applications. They are used in a variety of applications including diagnostic, therapeutic, and implantable devices (e.g. catheters [3], biosensors [4], artificial skin, controlled release drug delivery systems [5–11] and contact lenses [12]). Hydrogels have been widely used in such applications because of their compatibility with the human body and also because hydrogels resemble natural living tissue more than any other class of synthetic biomaterials. This is due to their high water content and soft consistency that is similar to natural tissue [13].

Selection of hydrogels used in such pharmaceutical processes depends on the characteristics of the gel and depending on the application of the drug or protein. Hydrogels have several important characteristics that play an important role in drug diffusion including ionisation of the gel, swelling ratio, and specific mesh or pore size. Functional groups along the polymer chain can also react to the external environment for example temperature [14–18], ionic strength [19–20], pH of the swelling agent [5, 21, 22], or a combination of two or more factors [23–27].

2. Experimental

2.1. Preparation of samples

The hydrogels investigated in this work were prepared by free-radical polymerisation. The monomers used were 1-vinyl-2-pyrrolidinone (NVP, Lancaster synthesis) and acrylic acid (AA, Merck-Schuchardt, Germany), these polymers were chemically crosslinked using ethylene glycol dimethacrylate (EGDMA) or polyethylene glycol dimethacrylate (PEG600DMA) crosslinking agents (Sigma Aldrich) at 0.1 wt% of the total monomer content. Both monomers and crosslinking agents were used as received.

For initial characterisation of the copolymers for their suitability for coating applications, 1-hydroxycyclohexylphenylketone (Irgacure® 184, Ciba speciality chemicals) was used as a UV-light sensitive initiator at 3 wt% of the total monomer weight. This was added to the NVP/AA monomeric mixture and stirred continuously until completely dissolved. The solution was pipetted into a silicone mould (W.P. Notcutt, Middlesex) that contained rectangular impressions. The mould was positioned horizontally to the gravity direction under two low intensity UVA 340 UV lamps (*Q*-panel products) and the solution was cured for 1–2 h in an enclosed environment. The samples were dried in a vacuum oven at 40°C, 260 mmHg for 24 h prior to use. These samples were used for swelling studies. The feed ratio of the monomers used was 70 wt% NVP and 30 wt%AA (designated 70–30) and 80 wt% NVP and 20 wt%AA (designated 80–20).

From the initial characterisation a monomeric concentration that showed the greatest potential was further developed for coating applications. To this monomeric concentration a photoinitiator that enabled grafting to occur was added at 3 wt% of the total monomer content. For drug release characterisation aspirin (Sigma Aldrich) was added to the monomeric concentration at 25 wt% of the total monomer content. The substrate used was a polyether block amide 'Pebax® 3533′, that had been extruded through a tape head die using a "Prism T20 U-Tron Soder" twin-screw extruder and the extrudate fed through a "Thermo Prism" haul-off mechanism.

2.2. Swelling studies

Swelling experiments were preformed in buffered solutions (buffered tablets, BDH Ltd. Poole, England) in triplicate at pH values of pH 4.0, pH 7.0 and pH 9.2 at ambient temperature. Samples of the cured polymer with an average mass of 1.44 g, were placed in a petri dish, and the petri dish was filled with 40 mL of the appropriate buffered solution. Periodically excess buffered solution was removed by pouring the solution through a Buchner funnel, the samples were then pat-dried with filter paper and weighed. The samples were re-submerged in fresh-buffered solution. Both the 'water content' (W_c) and 'water uptake' (W_u) of the fully hydrated hydrogels was calculated using the formulas:–

$$W_c = 100\frac{(W_t - W_o)}{W_t} \quad \text{and}$$

$$W_u = 100\frac{(W_t - W_o)}{W_o}$$

respectively; where W_t is the weight of the swollen hydrogel and W_o is the weight of the hydrogel before swelling experiments took place. This process was continued for approximately 120 h at which point the sample was removed from the buffered solution, weighed and dried in an oven at 80°C for 24 h.

2.3. Dip coating and UV curing

Before dip coating/UV curing was performed the substrate was hand-dipped into propan-1-ol to remove surface impurities, and then oven dried at 50°C for at least 30 min to evaporate the solvent.

Dip coating was carried out by immersing the substrate in a 100 mL-graduated cylinder that had being filled with the appropriate coating solution. The dipping speeds used were 22.1 ± 0.5 mm/s while dipping the sample, and 5.2 ± 0.035 mm/sec while withdrawing the sample. Immediately after dipping, the substrate was placed in the curing chamber, which consisted of two high intensity Dymax® 2000EC UV light sources, and cured for 120 s. These lamps were used as it would be impractical to attempt to cure a low viscosity coating on a substrate over a long period of time using low intensity UV lamps. When samples were to undergo more than one coating cycle, they were suspended from a retort stand for at least 5 min before a subsequent coating was applied. The samples were then oven dried at 50°C to remove any volatiles that may have remained after curing.

2.4. Optical microscopy

Optical analysis was carried out on samples that underwent a predetermined number of dipping/curing cycles to determine the optimal coverage possible. The test was carried out by swelling the pre-coated tape in a pH 7 buffered solution that contained 'Congo red' indicator, for 20 min. An 'Olympus BX60' microscope with a magnification of 10 X was used to characterise the coating at a microscopic level.

2.5. Coating characteristics

Coating characterisation was carried out on samples that had undergone 2 dip coating/UV curing cycles. The thickness of the coating was measured using a micrometer, capable of being read to 3 decimal places. The samples, whose thickness was to be measured, were dried for a period of 24 h at 50°C prior to being measured. The measurement was performed at three points along the length of 10 different samples, and the average thickness is illustrated in this work. This procedure was then repeated after the samples had being coated using the coating procedure described previously, and the coating thickness calculated.

The contact angle made between the coating solutions and the substrate was measured in triplicate using a goniometer.

The relative viscosity (η_{rel}) of the coatings was determined by placing 5 mL of a coating into a 50 mL volumetric flask, and then filling the volumetric flask to the mark with acetone. Using an Oswald viscometer the flow times for acetone and the coating solution was measured in triplicate. From these results the relative viscosity could be calculated using the formula:

$$\eta_{rel} = t_{sol}/t_0$$

where t_{sol} is the flow time for the coating solution, and t_0 is the flow time for the solvent.

The density of the coating solutions was determined by placing 5 mL of the coating into a pre-weighed beaker and determining the mass of the known volume of liquid. The density could then be calculated using the simple formula:–

$$\text{Density} = \text{mass/volume}$$

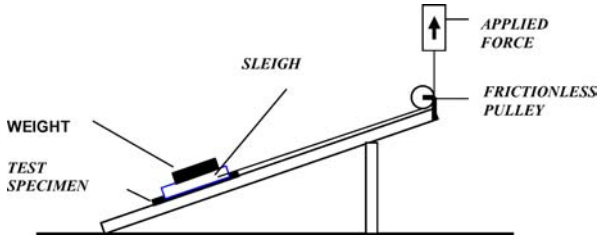

Figure 2 Setup of friction apparatus.

2.6. Fourier transform infrared spectroscopy

Fourier transform infrared spectroscopy was carried out on coated Pebax® 3533. The test was performed by running a background scan on uncoated Pebax® 3533, and thus the resultant scan was of the coating and not of the coating and the Pebax® 3533. The tests were carried out using a Nicolet Avator 360 Ftir, with a 32 scan per sample cycle.

2.7. Friction analysis

Samples of Pebax® 3533 that had an optimum number of coating cycles, were suspended in a beaker containing pH 7 buffered solution for varying amounts of time. At five minute time intervals a sample was removed and frictional analysis carried out. Frictional analysis was performed on the coating/glass interface, using a Lloyd LRX tensile testing machine with a 2.5 KN load cell. The test was carried out by placing the tape under a sled that was attached to the load cell as shown in Fig. 2. A force of 7.36 N was then placed on top of the sled. The sample and sled were then aligned with the crosshead of the testing apparatus, and the sample was pulled at an angle of 30° for a distance of 100 mm at a speed of 100 mm/min. From this test the optimal swelling time for the coating was established.

2.8. Aspirin release

For Aspirin release testing, 7 × 100 mm coated samples with and without aspirin incorporated, were placed into a 100 mL bottle that contained 100 mL of preheated (37°C) pH 7 buffered solution. The samples were placed in an oven at 37°C. Aspirin release was determined by performing UV spectroscopy on the swelling media, at predetermined time intervals using the coated samples without entrapped drug as a reference. The UV spectroscopy was performed using a Shimadzu UV 160, UV spectrometer.

3. Results and discussion

3.1. Preparation of samples

Crosslinked samples of NVP/AA were photopolymerised using Irgacure® 184 as a photoinitiator for initial characterisation of their suitability as a potential lubricious coating. These samples were cured on a silicone mould, and prior to use dried for 24 h in a vacuum

oven. From visual inspection of these samples there was no significant difference in the polymers prepared in this work and those prepared in previous work [28].

The Pebax® 3533 tape was extruded through a tape head die using a Prism T20 U-Tron Soder twin-screw extruder, the extrudate was then fed through a Thermo Prism haul-off mechanism. Rolls of tape were then stored in airtight bags until required.

These samples were cut into lengths of 150 ± 1 mm for dip coating. After the curing process the samples were cut to lengths of 100 ± 1 mm to remove any inconsistencies in the coating from the top and bottom of the tape caused by the method used to suspend the samples for dipping, or droplets that may have formed during the curing process.

3.2. Swelling studies

Swelling experiments were performed in solutions of various pH values. The swelling experiments were carried out by placing a circular disc of photopolymerised polymer in a petri dish. The polymer was then immersed in the appropriate buffered solution and allowed to swell. The amount that the hydrogel swelled was determined by periodically removing the buffered solution, pat-drying and weighing the sample.

The pK_{initial} of the copolymers tested in this work is in the range of 4.07 to 4.49 [28]. When the pH of the solution is less than the pK_{initial} there is no ionisation of the carboxylic acid groups, and hydrogen bonding is maximised. This causes the flexibility of the polymeric chains to be rather low. Carboxylic acid groups within the network ionise and attract cations into the gel to replace the H^+ ions as the pH of the environmental solution rises above its pK_{initial}. This effectively raises the concentration of free ions inside the gel causing the swelling to increase. Additionally, the gel tends to expand to minimise the repulsion between the ionised carboxylic acid groups [19].

All crosslinked samples had good gel integrity at the end of the swelling experiment. As expected the W_c and W_u (Fig. 3) values obtained increased as both the pH of the swelling media increased and the molecular weight of the crosslinking agents increased. The Wu results obtained by swelling both sets of gels were comparable in pH 4, with values of approximately 440 and 530 for the EGDMA crosslinked copolymers, and 912 and 940 for the PEG600DMA crosslinked copolymers for the 80–20 and 70–30 monomeric feed ratios respectively. However, as the pH of the swelling media increased there was a notable difference in the swelling characteristics of each set of samples, with the 70–30 monomeric concentrations yielding a W_u of up to 1300% more then 80–20 samples. This was due to the increased internal repulsion of the gel caused by the increased number of free ions liberated from the carboxylic acid groups of AA. The hydrogels were completely swollen, transparent, and had a smooth, continuous surface. They were also relatively firm but flexible. These observations are characteristic of a homogeneous crosslinked hydrogel that does not lose material by disentanglement of the polymer chains [23]. With both monomeric concentrations the polymers were observed to swell to a higher degree as the molecular weight of the crosslinking agents increased, due to additional free space between molecular chains.

It is also noteworthy that both hydrogels swelled to yield Wc values of over 90% when swollen in pH 7 or higher, this would imply that the hydrogels would be biocompatible as they resemble natural living tissue due to their high water content and soft consistency that is similar to natural tissue [13].

Swelling behaviour of a hydrogel differs depending on which step becomes dominant in determining the swelling rate. Depending on the dominant factor, the mechanism of transport for solvent penetration into the polymers can be classified as either Fickian (Case-I) diffusion or anomalous (non-Fickian, Case-II) transport. When no structural changes of the polymer network occur throughout the whole process or when the rate of diffusion or penetration is much less than that of the polymer hydration relaxation process, the penetration of solvent into the polymer is governed by solvent molecule diffusion through the polymer network (Step 1, Fig. 4). In the extreme case that the solvent diffusion rate is much faster than the relaxation rate, relaxation process becomes the rate-determining step (Step 2). When the gel expansion process dominates, the swelling kinetics is governed by collective diffusion (Step 3) [29] .

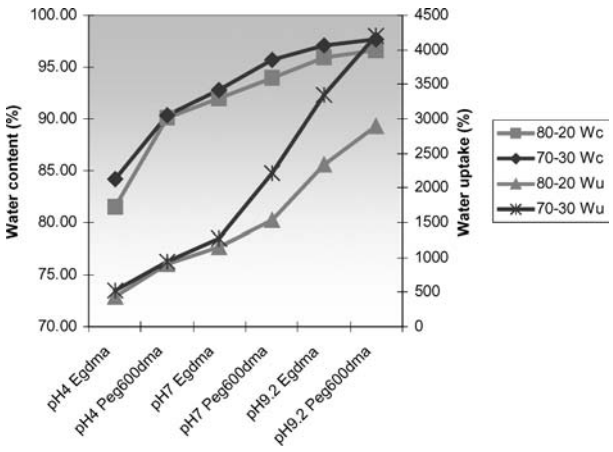

Figure 3 The percentage water content (W_c) and the percentage water uptake (W_u) of the hydrogels analysed.

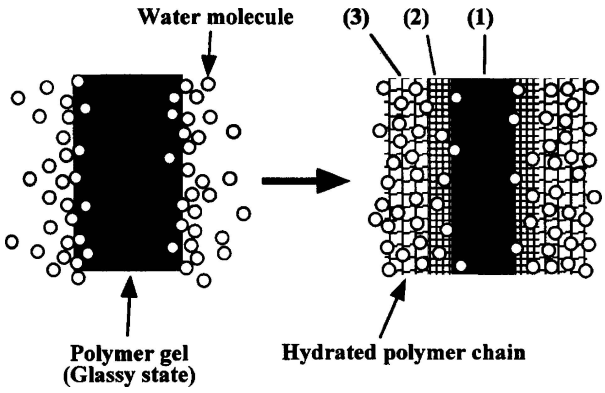

Figure 4 Water uptake process for polymer gels from initially glassy dry state [30].

3.3. Dip coating and UV curing

Dip coating was carried out by immersing the substrate into a 100 mL graduated cylinder that had being filled with the appropriate coating solution. Immediately after dipping, the substrate was placed on the curing apparatus, cured for 120 s and oven dried at 50°C for 24 h.

For coating purposes it was decided to use the 70–30 monomeric concentration as the monomeric feed ratio, as these samples swelled to a greater degree, which would be advantageous for coating applications. It was found that this coating wet the substrate during dip coating, and bonded readily during the curing process. The coating achieved on the Pebax® 3533 tape was slightly tacky after curing. However after drying there was no evidence of this tacky nature.

It was noteworthy that the stiffness of the tape increased significantly after coating, and as the coating was built up over a number of dipping/curing cycles the stiffness appeared to increase. This led to the coating cracking when the tape was bent, however the coating did not 'chip off'. This indicated grafting had taken place.

3.4. Optical microscopy

Optical analysis was carried out on samples that underwent a predetermined number of dipping/curing cycles to determine the optimal coverage possible. The test was carried out by firstly swelling the pre-coated tubing in a pH 7 buffered solution that contained Congo red indicator, for 20 min to hydrate the coating. An Olympus BX60 microscope with a magnification of 10 X was used to characterise the coating at a microscopic level.

In this work uncoated Pebax® 3533 was used as a reference for the coated samples. When the coated samples were examined on a microscopic level it was found that after 1 dipping/curing cycle, the coating achieved was inconsistent over the length of the sample, as the coverage achieved by the coating at the top of the samples was blotchy. However the coating achieved at the bottom of the tape appeared smooth and consistent. This was caused by the method of coating the substrate, as the bottom of the sample was in contact with the coating solution for a longer period of time than the top of the tape. Therefore the coating solution had more time to diffuse into the substrate and thus achieve better coverage of the substrate. It was for this reason that a faster dipping speed than withdrawal speed was used when dip coating, in an attempt to minimise this effect.

When the samples were dip coated a second time, it was found that the build-up of coating at the top of the samples greatly reduced the coverage problems observed after 1 dipping cycle. It was also noticeable that the coating at the bottom of the substrate was at least equal to the coating achieved after 1 dipping cycle. After 3 dipping cycles, it was found that both small and large clumps of coating were formed. This led to a poor surface finish being achieved, and a large difference in the coating along the length of the substrate.

When the samples that had undergone 4 dipping cycles were analysed, it was found that as well as there being both small clumps and large clumps present, there was also some very thick coating present. This also led to a poor surface finish being achieved, and a large difference in the coating along the length of the substrate. It is also noticeable that after 4 coating cycles some sections appeared to have little or no coating present. This was due to sections of coating that had being built up during previous dipping cycles, being effectively washed off by the final dipping cycle. These results illustrate that the best coverage by either coating tested, was achieved when the substrate underwent 2 dipping/curing cycles.

3.5. Coating characteristics

The thickness of the coating was measured using a micrometer, capable of being read to 3 decimal places. The sample thickness was measured at three points along its length, and the average taken as the thickness of the coating.

From this test it was found that the average thickness of the coating increased with the incorporation of aspirin, as shown in Table I. This could be caused by a lower polymerisation-induced shrinkage of the coating material, due to the incorporation of aspirin. This is justified in that the thickness variance in the coating that had the highest molecular weight crosslinking agent was greater than the variance in the coating that contained the lowest molecular weight crosslinking agent. There was also an increase observed in the standard deviation with the incorporation of aspirin, which suggested that the drug was not evenly distributed throughout the coating, and may have phase separated during polymerisation. However as the calculated drug content was between 63 and 80 mg for the EGDMA and PEG600DMA crosslinked polymer respectively, it would not be a hazard to any potential patient.

It was also found that there was an increase in the contact angle that the coating solution made with the substrate with the incorporation of aspirin. This led to less thinning of the coating as the substrate was lifted from the coating solution, due to the more adhesive interface. This would lead to a thicker coating being achieved on the substrate when the aspirin containing coating was used.

There was also a slight increase in both the viscosity and the density of the coatings with the incorporation of aspirin, which may also have contributed a thicker coating, been achieved when aspirin was incorporated into coating solution.

TABLE I Coating characteristics

	Crosslinking agent used			
	EGDMA		PEG600DMA	
Aspirin content (%)	0	25	0	25
Thickness (μm)	67.1	75.04	55.8	74.8
Std dev	0.0396	0.0495	0.0336	0.0681
Contact angle	36.167	41.33	38.83	42.67
Relative viscosity	1.212	1.235	1.212	1.242
Density	1.042	1.081	1.049	1.066

3.6. Fourier transform infrared spectroscopy

Fourier transform infrared spectroscopy was carried out on coated Pebax® 3533, in order to analysis the composition of the coating. The test was carried out by firstly performing a background on uncoated Pebax® 3533, and thus the resultant scan was of the coating.

From the literature the carbonyl group of PVP exhibits a peak between 1670–1680 cm^{-1} [5]. When the carbonyl group forms intermolecular bonding there is a negative shift exhibited in the IR spectrum. On analysis of the IR spectrum of the coating it was found that the PVP carbonyl group exhibited peaks in the region of 1660 to 1680 cm^{-1}, and at approximately 1630 cm^{-1}. This indicated that the crosslinking agent acted as a spacer that did not allow a high degree of intermolecular bonding to occur. Bures *et al.* [22] states that the carboxylic acid groups can exist in both free and dimeric form depending upon their environment. Lee *et al.* [31–32] describes how the stretching frequency of the carbonyl moiety in the carboxylic acid group as is defined by absorption at 1750 cm^{-1}, whereas the dimer stretching frequency has being reported as being between 1700–1720 cm^{-1} [5, 22, 28, 31, 32]. In the current study it was found that two shoulders were formed on the PVP carbonyl peak. The first shoulder was in the region 1720 cm^{-1}, which indicated the presence of the AA dimer. The second shoulder was in the region of 1740–1750 cm^{-1}, this in turn indicated non-hydrogen bonded AA.

It was also found that the peaks associated with C=C and the vinylic hydrogens at 1617 and 1409 cm^{-1} respectively were not present [33], thus proving that a very high degree of polymerisation had taken place. These results correlate with the Ftir results obtained for similar bulk copolymers tested in previous work [34], and thus prove that the application of this copolymer to a substrate for coating applications does not affect its chemical structure.

3.7. Friction analysis

Friction analysis was carried out on samples to characterise the lubricious nature of the hydrogels that contained either EGDMA or PEG600DMA as a crosslinking agent. This test was performed on samples that had undergone two coating cycles, as this had being proven to give the optimum coverage. The dry, coated reference samples were dried in an oven at 50°C for 24 h prior to testing to remove any moisture that may have being absorbed.

From the static coefficient illustrated in Fig. 5 for the coatings that contained EGDMA and PEG600DMA respectively, it can clearly be seen that the static coefficient of friction, for the coated samples whose coating was fully hydrated was below 0.2. This value was significantly less than the value achieved for the uncoated substrate (1.12). Thus proving the lubricious nature of the coating. LaPorte [35] quoted values of 0.74 for the static coefficient of HDPE from the sales literature of Boston Scientific Corporation. This value was reduced to 0.38 when a silicone oil-based coating was used on

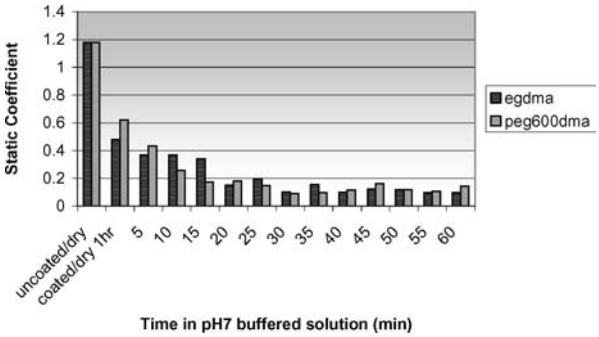

Figure 5 Coefficient of friction results for samples crosslinked with EGDMA and Peg600DMA.

the substrate, and 0.05 when a hydrophilic coating was used. The values found in this work were half that of the values quoted for the silicone oil-based coating and would not have the handling problems associated with this type of coating [2]. However the values achieved were higher than the values quoted for the hydrophilic coating. Nevertheless it is noteworthy that in this work the coefficients of friction quoted are frictional results between the coating being tested and a glass fixed plane at an angle of 30°. As friction is a characteristic of the surfaces tested the values achieved cannot be compared directly to those quoted by LaPorte as the test conditions for the quoted results are not clear. It should also be said that the values achieved in this work are below 0.2 and in many cases in the region of 0.1.

It was also found that complete hydration of the coatings tested differed with the molecular weight of the crosslinking agent used. In essence when EGDMA was the crosslinking agent used, complete hydration did not occur until the coating was tested after 20 min. When PEG600DMA was used to crosslink the copolymer complete hydration had almost been achieved after 10 min. This correlates with results from swelling studies, where it was found that the Wc and the Wu increased as the molecular weight of the crosslinking agent was increased. This was due to an increase in free space in the coating into which water could diffuse, thus the polymer could swell more and as a coating, swell faster. Even though the coating did not swell instantly, it does compare favourably with swelling rates found by Nurdin *et al.* [36]. Their work involved the analysis of a commercially available catheter coating, Hydromer®. It was found that after several hours in water, the surface was totally wetted by water, and suggested that pre-hydration may be necessary.

3.8. Aspirin release

Aspirin release was determined by performing UV spectroscopy using a Shimadzu UV 160 UV spectrometer. A sample of the swelling media was periodically tested to determine the Aspirin release profile.

From the release profile shown in Fig. 6 it can be see that the release profile can be broken down into three almost linear regions. The first of these regions was caused by the release of aspirin close to or on the surface of the coating. This release took place within the first 15 min of the test, where 60% of the total drug content

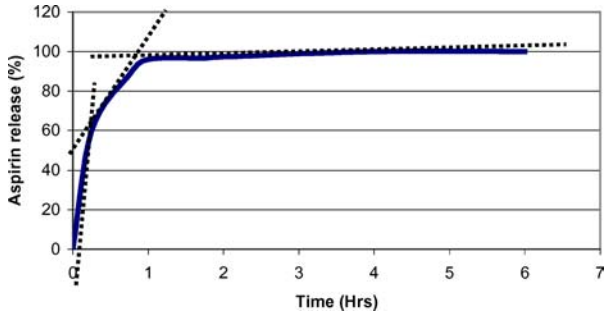

Figure 6 Release profile for NVP/AA co-polymer crosslinked with EGDMA crosslinking agent.

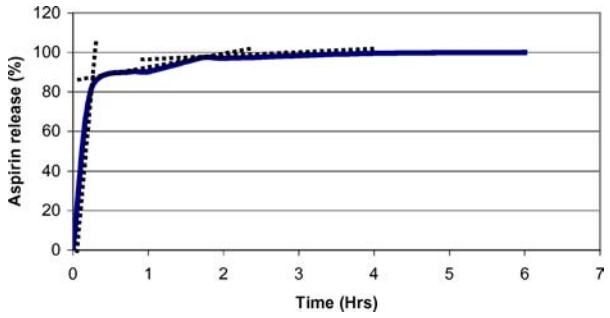

Figure 7 Release profile for NVP/AA co-polymer crosslinked with PEG600DMA crosslinking agent.

was released. This release was replaced by a slower release rate that consisted of the release of aspirin that was entrapped within the bulk of the coating, which continued for approximately 60 min. Finally the third and slowest release rate observed was the release of any remaining drug.

The release profile achieved for the co-polymer crosslinked with PEG600DMA (Fig. 7) yielded a similar release profile. However, as the crosslinks in the polymer were longer, the polymer swelled more quickly and thus allowed drug close to the surface to be released, which resulted in 80% of the total drug content been released in the initial release phase. The second release phase was also extended in relation to the EGDMA crosslinked coating, and lasted up to 2 h. However most of the drug had already been released prior to this release phase.

It is noteworthy that the drug content was slightly higher in the EGDMA crosslinked copolymer (80 mg) in comparison to the PEG600DMA (63 mg) crosslinked copolymer. The drug content was calculated using the dimensions of the tape prior to coating, the coating thickness, total mass of the samples after coating, as well as the density of Pebax® 3533 [msds sheet]. This correlates with friction results in that full hydration of the PEG600DMA crosslinked copolymer occurred more rapidly than the EGDMA crosslinked copolymer, and therefore the buffered solution could penetrate the hydrogel structure quicker and allow the aspirin to diffuse out.

4. Conclusions

In this work we have developed a lubricious hydrophilic coating that had potential applications in the medical device industry. We have shown that increasing the thickness of a traditional solvent based hydrophilic coating by the polymerisation of monomers directly onto the surface of the substrate is both an effective and simple method of encapsulating a drug within a coating, and eliminates the use of potentially hazardous solvents.

Even though the samples did appear tacky after curing, any remaining monomers were removed by oven drying the samples. This was proven using Ftir analysis where no characteristic monomeric peaks were found.

Optical microscopy suggested that 2 coating cycles yielded the most consistent coating throughout the entire length of the substrate. Frictional analysis proved the lubricious nature of the coating, even though it was slightly thicker than solvent dispersed coatings. Finally, drug release analysis showed that the release of an active agent could be controlled by varying the molecular weight of the crosslinking agent utilised.

In conclusion this coating has potential to coat several different substrates, as the coating has bonded readily to stainless steel in initial trials. The coating also has the potential to release several active agents, due to the method of incorporation and release. Therefore, the lubricious hydrophilic coating developed in this work has potential applications in the medical device industry.

Acknowledgments

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