al. 2000). When there was a move towards sustained release systems natural polymers were considered for their production such as albumin, gelatin, collagen and alginate but the favour for these polymers has waned over the years due to batch inconsistency and a contamination hazard due to prions (Matschke et al. 2002). With the emergence of *in situ* forming systems, synthetic polymers were considered for use. The most commonly employed polymers for ISFI are of a synthetic nature (Abashzadeh et al. 2011; Astaneh et al. 2009) and as previously mentioned are usually biodegradable and biocompatible (Hatefi and Amsden 2002; Luan and Bodmeier 2006). These polymers are hydrophobic and insoluble in water and it is this characteristic that allows for the solid implant to form after polymer precipitation takes place (Hatefi and Amsden 2002; Brodbeck et al. 1999; Liu et al. 2010). Polymer selection for industrial production also involves consideration of both chemical and physical stability therefore careful selection of polymers is required (Matschke et al. 2002).

Biodegradable polymers have risen in popularity as they have a distinct advantage over those that are non-biodegradable. Non-biodegradable systems require surgery to implant the system and once the drug supply has been spent, invasive surgery is necessary to remove the implant from its site of injection (Lee et al. 2010; DesNoyer and McHugh 2001). Regarding the use of non-biodegradable polymers in the treatment of vitreo-retinal diseases, this invasive surgery has been linked with a number of serious side effects such as vitreous haemorrhage, cataract formation and retinal detachment (Kiernan and Mieler 2009; Yasukawa et al. 2001; Mohammad et al. 2007). With biodegradable polymers, invasive surgery is avoided as the implant degrades to form non-toxic by products (Luan and Bodmeier 2006), which are removed by the tricarboxylic acid cycle (Krebs cycle) that yields carbon dioxide and water as metabolic end products (Schoenhammer et al. 2010).

Biodegradable polymers that are commonly used in SPI systems are from polyhydroxyacid, polyanhydride and polyorthoester families. Aliphatic esters from the poly-α-hydroxyacid family such as poly(glycolic acid) (PGA), poly(lactic acid) (PLA) and PLGA which is a co-polymer of PGA and PLA, are extremely popular (Brannon-Peppas 1995; Lee et al. 2010). Poly-ε-caprolactone (PCL) (Bae et al. 2006), poly(lactide-co-caprolactone) copolymer, poly(acrylic acid) (PAA) and its derivatives (Haglund et al. 1996) such as poly(methacrylic acid) (PMA)—poly(ethylene glycol) (PEG) have also being investigated as potential SPI polymers (Ismail et al. 2000). These polyesters degrade by random chain scissions of ester bonds that result in a steady reduction in molecular weight but a delayed loss of weight, known as bulk hydrolysis.

PLA and PLGA have been the most popular polymers in SPI formulation (Table 1). PLGA has a long history of use in biomedical applications and was described in the earliest work completed by Dunn et al. 1990. It has been approved for parenteral use (Eliaz and Kost 2000; Fredenberg et al. 2011) by US Food and Drug Administration (US-FDA), which is prepared by polymerisation of lactic acid and glycolic acid monomers (Merkli et al. 1998). The glass transition temperature (T_g) of PLGA copolymers are above physiological temperature of 37ºC, which imparts a moderately rigid chain configuration and therefore the mechanical strength at ambient temperatures. Jamshidi et al. 1988 observed a decrease in T_g when the ratio of lactic acid monomers also decreased. Availability of PLGA in different commercial grades such as lactide to glycolide ratio and molecular weight is also raised its popularity, as