to allow a continuous therapeutic action in situ with reduced toxicity. From a general viewpoint, the treatment of many pathological conditions (including osteomyelitis) can potentially benefit from local and temporally controlled administration of pharmaceutical agents. In this regard, bioactive glasses have been widely investigated as carriers of a number of drugs to enhance their therapeutic and tissue regeneration potential (Hum and Boccaccini, 2012). Drugs and growth factors cannot be incorporated into glass during the manufacturing process if the material is produced by a melting route due to thermal degradation of biomolecules. However, they can be mixed into the sol suspension if the glass is synthesized by a low-temperature sol-gel method (Andrade et al., 2009; Cevc and Blume, 2004). Another approach involves the incorporation of the therapeutic molecules into the organic phase of polymer/bioactive glass composites (Habraken et al., 2007). However, both these approaches have disadvantages. It is not possible to apply a high-temperature treatment to stabilize sol-gel materials with embedded drugs or for the removal of organic by-products. Similarly, complex procedures to fabricate drug-containing composites may not be possible.

Once the glass has been produced and no additional thermal treatment is required, postsynthesis incorporation of biomolecules into the glass seems to be the most feasible strategy. Hence, silicate mesoporous materials have attracted significant attention from biomedical researchers due to their potential of acting as an efficient platform for controlled drug release (Mamaeva et al., 2013; Arcos and Vallet-Regí, 2013). Pure SiO_2 mesoporous materials were first synthesized in the early 1990s by researchers of the Mobil Oil Corporation (Kresge et al., 1992) and, about one decade later, Yan et al. (2004) pioneered the development of SiO_2-CaO-P_2O_5 mesoporous bioactive glasses (MBGs) that successfully combined bone-bonding and drug-release capabilities. The synthesis of MBGs is typically based on a cooperative self-assembly of supramolecular surfactant molecules, acting as structure directing agents, and oligomeric silica species that form the glass network. MBGs are biocompatible and have a highly ordered mesoporous structure, with a channel diameter of between 5 and 20 nm (Arcos and Vallet-Regí, 2013). Compared with conventional melt-derived (nonmesoporous) bioactive glasses, MBGs possess a significantly higher specific surface area (well above 100 m^2/g compared with $<1\ m^2/g$), and hence exhibit faster kinetics of apatite formation on their surface in vitro (mineralization) (Wu and Chang, 2012). The high pore volume and surface area permit high loading efficiency of biomolecules into the mesopores as well as slow and controllable drug-release kinetics compared with conventional nonporous melt-derived glasses. The formation of a hydroxyapatite-like layer on the walls of mesopores indicates a partial occlusion of the channels, which decreases the burst release effect and the overall release rate, thereby allowing a prolonged therapeutic effect to be obtained (Wu et al., 2010a). Table 2.1 provides an overview of the various growth factors and drugs that have been incorporated into the mesopores of MBG-based systems.