co-surfactants contribute to the formation of disordered mesopores after the removal of organic templates. The rare-earth ions may affect their porous structure (Huang et al., 2012). The MBG:$Eu^{3+}$ nanofibers have a surface area of 188 $m^2$/g, a pore volume of 0.246 $cm^3$/g, and an average pore size of 4.17 nm, while the MBG:$Tb^{3+}$ nanofibers have a surface area of 171 $m^2$/g, a pore volume of 0.186 $cm^3$/g, and average pore size of 3.65 nm (Huang et al., 2012). The great surface areas and mesoporous channels make them appropriate as drug delivery systems with excellent drug-loading and controlled release properties for ibuprofen (IBU). Interestingly, the emission intensities of $Eu^{3+}$ in the drug delivery system vary with the released amount of IBU, thus allowing the drug release to be easily tracked and monitored by the change in luminescence intensity (Huang et al., 2012).

In order to increase the surface areas and drug delivery properties, MBG hollow fibers were fabricated by an electrospinning technique using poly(ethylene oxide) (PEO) as a phase separation agent (Hong et al., 2010). The formation mechanism of hollow cores in the MBG fibers is attributed to the microphase separation between the volatile solvent and nonvolatile components induced by rapid solvent evaporation during electrospinning. In the above system, the water/ethanol mixing solvents play an important role. When the low water/ethanol ratio is 0.5, thick-walled bamboo-like hollow fibers are produced. The increase in the water/ethanol ratio widens the diameter of the hollow core and thins the wall (Hong et al., 2010). The drug-loading capacity and drug-release performance of the MBG hollow fibers depend mainly on the fiber length. If the fiber length is >50 μm, the MBG hollow fibers become excellent carriers for drug delivery. The shortening of their fiber lengths may reduce the drug-loading amounts and accelerates drug release (Hong et al., 2010).

### 4.2.4 MBG Coatings

Metals and alloys have been widely used in orthopedic and dental fields under load-bearing conditions due to their low density, good corrosion resistance, and excellent mechanical properties (Paital and Dahotre, 2009). However, the bio-inert property limits their clinical applications. The above disadvantage is overcome by the deposition of the MBG coatings on medical metals because they combine the mechanical advantages of metal alloys with good biological properties of MBGs (Shruti et al., 2016; Zhang et al., 2016b; Li et al., 2015b; Wang and Wen, 2014; Ye et al., 2014; Huang et al., 2014, 2013; Wang et al., 2010; Gomez-Vega et al., 2001). The MBG coatings have the following advantages for bone implants: (i) they serve as a shield to avoid toxic ions released from metal implants that are harmful for human cells under high ion concentrations; (ii) the coatings provide compatible implant-tissue interfaces, and thus enhance the osseointegration between implants and bone tissues; and (iii) the MBG coatings on magnesium alloys not only accelerate the apatite deposition, but also act as a barrier to decrease the degradation rate of magnesium alloys (Li et al., 2015b; Huang et al., 2014).