pathogen disease transfer from the donor to the patient, and problems concerning the graft quality can occur (Ries et al., 1994).

The second strategy to replace bone losses is the use of man-made materials as implants to interface with living tissues, in order to support the biomechanical loads during the bone regeneration process (Jones and Hench, 2003).

Since the early 1970s, bioceramics have been extensively proposed by many researchers for bone repair. Natural or synthetic hydroxyapatite has been used for bone replacement because of its chemical and crystallographic similarity to the carbonated apatite in human teeth and bone (Ozawa et al., 1989; Thomson et al., 1998). Other calcium phosphate materials, such as β-tricalcium phosphate (β-TCP), can act as hydroxyapatite precursors and have been successfully proposed in the form of particles for filling small bone defects in orthopedics and dentistry (LeGeros, 1993; LeGeros et al., 2003; Xigeng et al., 2008). Bioactive glasses and glass-ceramics have been widely studied due to their remarkable ability to bond to bone and to stimulate the growth of new bone (Hench et al., 1972; Hench, 1998a,b). Recent findings indicate that controlled release of ionic dissolution products from bioactive glasses can be used to induce angiogenesis and thereby offer a great potential for design of gene activating glasses for tissue regeneration. In fact, it was demonstrated that biologically active Si and Ca ions can stimulate the genes of the cells toward a path of self-repair (Hench, 2009), and other trace elements can be exploited to promote angiogenesis (e.g., $Co^{2+}$ (Wu et al., 2012; Kargozar et al., 2017)) or therapeutic functions (e.g., antibacterial effect through $Ag^{+}$ release (Miola et al., 2016)).

Bioactive glasses have been often used to produce scaffolds for bone regeneration (Baino and Vitale-Brovarone, 2011; Fu et al., 2011; Baino et al., 2015). The criteria that an ideal scaffold should fulfill have been described by many authors (Adachi et al., 2006; Livingston et al., 2002; De Aza et al., 2003; Hutmacher, 2000; Kieswetter et al., 1996). In general, a scaffold for bone tissue engineering should (i) be biocompatible, producing nontoxic products of degradation, (ii) exhibit a 3D structure of highly interconnected macropores (above 50 vol.% like in human cancellous bone), (iii) have large macropores (100–500 μm) to allow cell access, new bone formation, and blood vessel growth, (iv) promote cell adhesion, spreading proliferation and differentiation, (v) have mechanical properties matching those of the host bone, and (vi) be easily shapable and processable. A number of methods have been experimented with to produce bioceramic and bioactive glass scaffolds; essentially, the porosity can be introduced by using three major approaches, that is, burn off of a polymeric phase (e.g., starch (Lyckfeldt and Ferreira, 1998), polyethylene spheres (Baino et al., 2009), or a porous sponge (Baino et al., 2016a)), foaming strategies (Jones et al., 2006; Midha et al., 2013), or additive manufacturing (Gmeiner et al., 2015).

As reviewed by Miao and Sun (Miao and Sun, 2010), there is quite an abundant literature on the use of hydroxyapatite and other calcium-phosphate ceramics, along with composites with biocompatible polymers, for making