Introduction

Articular cartilage is a thin layer of connective tissue located within joints and on the ends of the long bones (Buckwalter and Mankin, 1997). Articular cartilage functions as a low-friction, wear-resistant, load-bearing material that facilitates joint motion (Maroudas, 1979; Mow and Ratcliffe, 1997). The articular cartilage of diarthrodial joints experiences a high level of biomechanical stress over many decades (Hodge et al., 1986) and, in many cases, can tolerate years of repetitive loading. However, cartilage damage and degeneration occur often with traumatic joint injury and advancing age at particular sites, such as the knee and hip.

The poor intrinsic healing capacity of articular cartilage has been described for centuries, and continues to be a problem today (Buckwalter and Mankin, 1998). This inadequate healing response is likely related to the relatively low cellularity and low metabolic activity of the chondrocytes within the tissue. Additionally, because of the avascularity of cartilage, cells that participate in systemic wound-healing responses are not able to access the injury site and promote tissue repair. When cartilage damage penetrates the subchondral bone, however, a cell-based repair response is initiated. Nevertheless, such a response results in the generation of a fibrocartilaginous repair tissue that does not match the composition, structure, or function of healthy articular cartilage.

The attainment of a number of specific design goals is likely to be critical to the development of a consistently successful strategy for the repair of cartilage defects (Hjertquist and Lemperg, 1969; Sah et al., 2001). One goal is to ultimately form tissue that has the normal site-specific biomechanical characteristics of articular cartilage and subchondral bone. A second goal is that the formed tissue should integrate firmly to both the adjacent host cartilage and the underlying bone. To attain these goals, biological processes of repair and regeneration need to occur under the influence of postoperative biomechanical demands, which can be affected markedly by rehabilitation regimens. Thus, the time-dependent processes of cell-mediated matrix remodeling and matrix-dependent alteration of cartilage biomechanical properties need to be determined and controlled.

Current clinical strategies for treating defects of eroded or degenerated cartilage often involve the surgical clearing of the affected tissue area followed by the implantation of grafts that replace or form cartilage or, alternatively, both cartilage and bone. At one extreme is the approach of implanting preformed cartilage and bone; for example, an osteochondral fragment can be transplanted from an allogenic (Fig. 15.1A) or autogenic (Fig. 15.1B) source. At the other extreme is the approach of filling the bulk of the defect with cells that can facilitate the growth of appropriate cartilage and/or bone tissue. An example of this is autologous chondrocyte implantation (Fig. 15.1C), wherein a high-density cell suspension is introduced into the defect site under a tissue flap, and cartilaginous repair tissue is formed (Brittberg et al., 1994;
Grande et al., 1987). Experimentally, growth factors have been applied in order to stimulate the metabolic activity of cells in a repair situation (Fujimoto et al., 1999; Nixon et al., 1999). In between these extremes are the experimental strategies of forming immature cell- and matrix-laden tissue constructs in vitro and implanting the constructs into defects in vivo, after which maturation occurs.

The United States National Committee on Biomechanics goals for promoting functional tissue engineering include the identification of structural and mechanical requirements for engineered tissue constructs (Butler et al., 2000). To achieve this goal, in vivo mechanical loads need to be assessed in order to characterize the environment that a tissue-engineered construct will experience upon implantation. Patient-specific factors, such as age, health, and expectation for future use, may be key determinants of this mechanical loading environment. In addition, the mechanical properties of the tissue, whether a construct fabricated in vitro or a tissue formed in vivo, must be determined within the framework of a mechanical model. Perhaps most importantly, the biological response and adaptation of the construct to the in vivo environment must be determined. An implant primarily composed of cells and relatively little matrix may evolve into functional tissue if it is mechanically protected during the early stages of growth, but may fail if overloaded. On the other hand, an implant that is composed of fully functional cell-laden cartilage may require early motion to maintain tissue homeostasis and attain a successful outcome.

Growth, resorption, and remodeling are fundamental processes that influence the size, shape, and properties of biological organs and tissues. Growth is “a normal process of increase in size of an organism as a result of accretion of tissue similar to that originally present” (Dorland, 1981). Here, volumetric growth of a constituent is interpreted as the deposition of constituent mass that has the same mechanical properties as the existing material. Volumetric