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Computational analyses of different intervertebral cages for lumbar spinal fusion

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ABSTRACT

Lumbar spinal fusion is the most common approach for treating spinal disorders such as degeneration or instability. Although this procedure has been performed for many years, there are still important challenges that must be overcome and questions that need to be addressed regarding the high rates of non-union. The present finite element model study aimed to investigate the influence of different cage designs on the fusion process.

An axisymmetric finite element model of a spinal segment with an interbody fusion cage was used. The fusion process was based on an existing mechano-regulation algorithm for tissue formation. With this model, the following principal concepts of cage design were investigated: (1) different cage geometries with constant compressive stiffness and (2) cage designs optimized to provide the ideal mechanical stimulus for bone formation, first at the beginning of fusion and then throughout the entire fusion process.

The cage geometry substantially influenced the fusion outcome. A cage that created an optimized *initial* mechanical stimulus did not necessarily lead to accelerated fusion, but rather resulted in delayed fusion or non-union. In contrast, a cage made of a degradable material produced a significantly higher amount of bone and resulted in higher segmental stiffness. However, different compressive loads (250, 500 and 1000 N) substantially affected the amount of newly formed bone tissue.

The results of the present study suggest that aiming for an optimal initial mechanical stimulus may be misleading because the initial mechanical environment is not preserved throughout the bone modeling process.

1. Introduction

Lumbar spinal fusion is currently the most effective surgical treatment in patients with spinal deformity, degenerative disk diseases, and degenerative spondylolisthesis. However, none of the existing clinical devices is completely successful. Non-union rates between 10 and 40% for mono-segmental treatment and, more frequently, for multi-segmental treatment have been reported in the literature (Blumenthal and Ohnmeiss, 2003; Bridwell et al., 1993; Fischgrund et al., 1997; France et al., 1999; Kuslich et al., 1998). One mechanical reason for this high rate of non-union is related to stress-shielding effects, i.e., inappropriate load sharing between an implant and the surrounding tissue, leading to a reduction in the mechanical stimulus required for bone formation (Huiskes et al., 1992; Otani et al., 1993).

Both the cage geometry and the material properties strongly influence the fusion outcome. In particular, clinical studies have suggested that polyetheretherketone (PEEK) cages show higher fusion and lower complication rates than titanium cages do (Chou et al., 2008; Niu et al., 2010). In a sheep study, cylinder-shaped cages resulted in accelerated fusion compared with box-shaped cages (Kandziora et al., 2002). *In vitro* studies also indicated that a larger, contiguous hollow space is important for decreasing the stress-shielding effect on a graft within an interbody fusion device (Epari et al., 2005; Kanayama et al., 2000).

Although mechano-biological finite element (FE) models of fracture healing in long bones are well established (e.g., Isaksson et al. (2008)), an iterative computational approach was only recently employed to investigate tissue formation during the fusion process (Postigo et al., 2014). These preliminary results showed that different rules for the mechanical regulation of tissue

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differentiation (Claes and Heigele, 1999; Lacroix and Prendergast, 2002) result in similar fusion patterns.

Several modeling studies on osteosynthesis devices reported a reduced healing time for long-bone fractures after optimization of the fixation device (Nolte et al., 2007; Wehner et al., 2012). In spinal fusion, numerical optimization has been used to design spinal cages to provide sufficient primary stability and compliance to avoid stress shielding (Kang et al., 2013; Lin et al., 2007, 2004). However, the potential for cage design optimization based on a mechano-regulation model has not yet been investigated. Therefore, the present study aimed to investigate bone formation during the fusion process with different cage designs. Particular efforts were made to examine whether optimization of a cage design based solely on the mechanical environment immediately after implantation is a viable approach for improving the entire fusion outcome. We hypothesized that this design causes accelerated bone formation in the early fusion phase but fails to create solid bone bridging in the later fusion phase. In contrast, bone formation within the entire intervertebral space should be achieved when the cage is optimized throughout the entire fusion process.

2. Materials and methods

2.1. Finite element model

An axisymmetric FE model consisting of two lumbar vertebral bodies, an interbody fusion cage and a fusion zone was used (Postigo et al., 2014) (Fig. 1a). The vertebral bodies represented the average geometry of an L4–L5 spinal motion segment (Panjabi et al., 1992). The fusion zone was defined as being inside the intervertebral space as well as within a 10 mm band around the circumference of the vertebral bodies. Each vertebral body contained the cortical shell and the cancellous bone. Eight-node fully integrated elements with quadratic displacement fields were used. The element size was chosen based on a mesh refinement study in which differences in stresses and strains with higher-density meshes were less than 3%. All material properties were linear elastic, isotropic, and homogeneous (Table 1). The FE software ABAQUS 6.10 (Simulia, Providence, RI, USA) was used for the simulations.

2.2. Bone fusion algorithm

An iterative procedure similar to that described by Lacroix and Prendergast (2002) for the mechano-biological tissue differentiation of long bone fracture healing was used (Fig. 1b). It was assumed that the entire fusion region was initially filled with granulation tissue. Precursor cell migration from the bone marrow into the fusion region was modeled as a diffusive process, with

$$\mathrm{d}n/\mathrm{d}t = D\nabla^2 n \tag{1}$$

where the cell density n is determined from the diffusion coefficient D, which is specifically chosen to reach the maximal cell density in the whole intervertebral space within the first 28 iterations. The cell origin was modeled as a constant concentration at the vertebral endplates, whereas the maximal cell density was normalized to 1. For simplicity, D was assumed to be constant, thus allowing for the calculation of cell migration in a separate analysis outside the iterative cycle. Furthermore, neovascularization was assumed to occur quickly compared with bone formation and was not explicitly modeled.

In each iteration, the mechanical stimulus (distortional equivalent strain ε_d and hydrostatic pressure σ_H, eq. (2)) was calculated according to the mechano-regulation theory proposed by Claes and Heigele (1999):

$$\varepsilon_d = \frac{2}{3}\sqrt{\left[(\varepsilon_1-\varepsilon_2)^2+(\varepsilon_2-\varepsilon_3)^2+(\varepsilon_1-\varepsilon_3)^2\right]},\ \sigma_H = \frac{1}{3}\mathrm{tr}(\underline{\sigma}) \tag{2}$$

where $\varepsilon 1$, $\varepsilon 2$ and $\varepsilon 3$ are principal strains and σ is the stress tensor. The original threshold values of ± 0.15 MPa for hydrostatic pressure and 5% for equivalent strain were used. The resorption limits

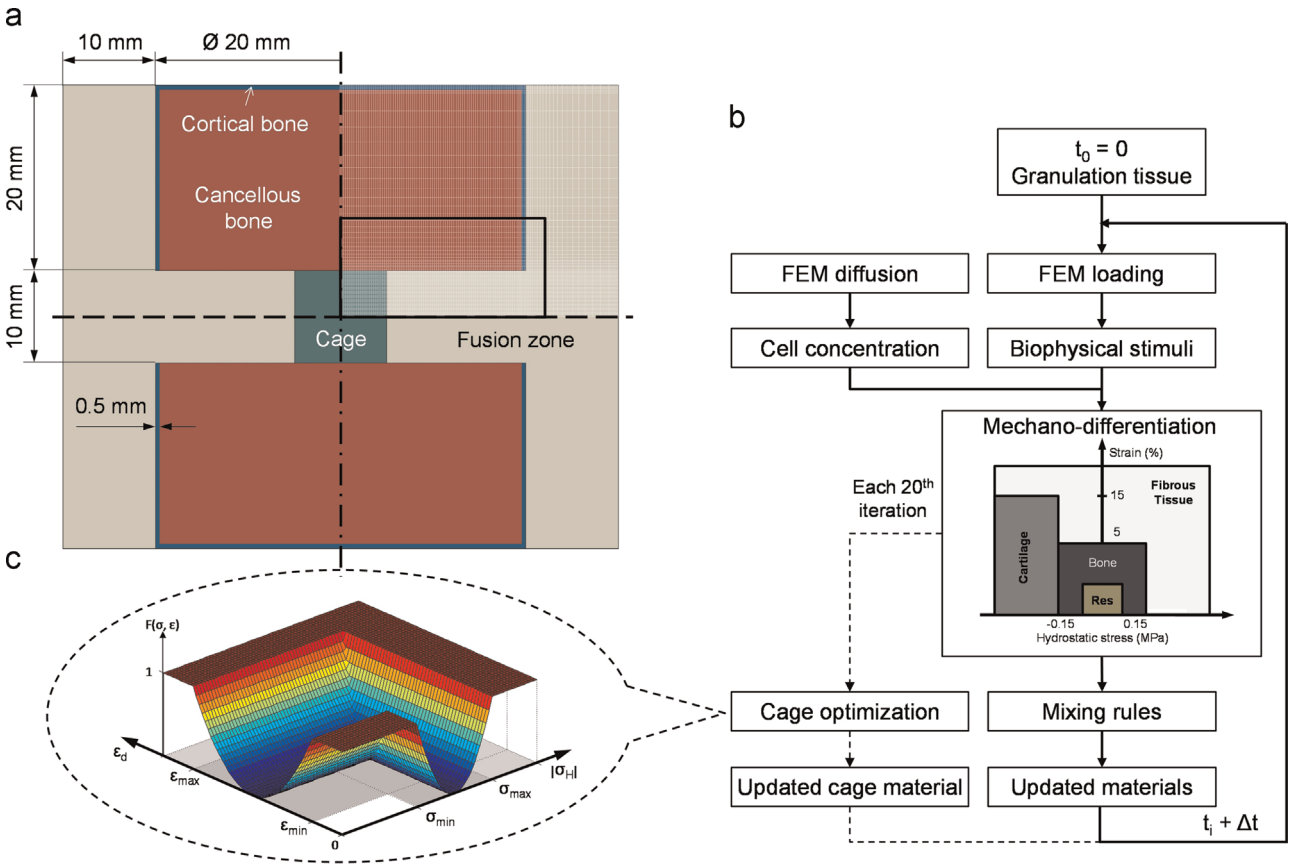

Fig. 1. (a) Axisymmetric finite element model of the spinal fusion region within two adjacent lumbar vertebral bodies and a centrally placed cylindrical fusion cage. The black rectangle represents the boundary of the bone density plots shown later in the results. (b) Flow chart of the bone remodeling process used. The mechano-differential algorithm was based on the theory of Claes and Heigele (1999). (c) Three-dimensional plot of the objective function, used for the optimization of the cage design.