MODELLING AND SIMULATION OF GROWTH OF SMOOTH MUSCLE CELLS WITHIN A CELLULAR SCAFFOLD OF POLY(CARBONATE-UREA)URETHANE AND THEIR EFFECT ON MECHANICAL PROPERTIES

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BACKGROUND

Cardiovascular diseases, (typically arteriosclerosis), affects a large population in the UK and the USA, and ischaemic heart disease with its lethal complication, the myocardial infarction, (heart attack), is the leading cause of death in these countries.

During reparative surgery, grafts are used to supplement diseased or blocked arteries but the patency of these grafts in lower limb and coronary applications is very poor due to the inherent lack of elasticity and formation of scar tissue, termed intimal hyperplasia (IH). This is caused by the uncontrolled proliferation of smooth muscle cells (SMC) in the damaged endothelium at the anastomosis.

Using a new type of polyurethane chemistry – poly(carbonate-urea)urethane which is highly biostable and biocompatible, [1,2], a new type of graft has been developed, similar to that shown in Figure 1. Unlike other polyurethane grafts it achieves its viscoelasticity during pulsatile flow by a process of wall compression in its honeycomb structure without the need for external dilation.

However these grafts are intended for long term use and their structure encourages the deposition and infiltration of new smooth muscle cells and coated at the lumen with endothelia, [3]. The mechanical response of any material used as the scaffold for the matrix of smooth muscle cells will therefore change, dependent on the mechanical properties and density of the infiltrated cells and the relative number and elasticity remaining pores. As a result of the highly non-linear material behaviour, inhomogeneous properties and geometry, it is not possible to find closed form analytical solutions.

COMPUTATIONAL METHODOLOGY

This paper reports on the application of a methodology using the Finite Volume Method, [4] which has been adapted to model the change to the overall elasticity and its effect on compliance due to the deposition and infiltration of new smooth muscle cells coated with endothelia.

This work unified the modelling of many different types of cellular solids, [5] and resolved seeding strategies to determine ultimate infiltration depth of the cells, their initial and progressed density, their stiffness, and the depth and variability of pore size of the material.

During the course of the work, the diffusion model was extended to include a simple dynamic random seeded cell-growth capability to analyse probable cell replication in an in vitro environment based on a novel experiment using a gel support matrix-scaffold enabled elementary validation of a complete 3d-model conducted earlier, [6]. This will assist in the quantification of quality control systems necessary to support the ongoing development in clinical trials particularly across a range of disease levels or cell division capability.

The finite element method, (FEM) is commonly used to predict stress and strain as a result of thermal events or physical loads, [7]. However, the finite volume method, (FVM) introduced in thermodynamics under the 'control-volume' name in the early 1970's, a special form of the finite difference method, (FDM) is now used extensively in turbulence modelling and other related fluid flow problems in Computational Fluid Dynamics, (CFD). More recently, the technique has been applied to transport in solid materials to good effect particularly in the analysis and simulation of thermal events and this has recently been translated for use in medical sciences.

The control volume integration results in statements which express the exact conservation of relevant properties for each finite cell. This clear relationship between the numerical algorithm and the underlying physical conservation principle forms one of the main attractions of the finite volume method and makes its concepts much more simple to understand than finite element methods.

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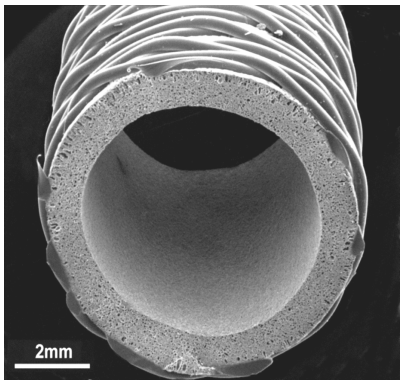

Fig. 1. *Externally reinforced MyoLink graft*