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A 3-D FINITE ELEMENT MODEL OF BLOOD PERFUSED RAT GASTROCNEMIUS MEDIALIS MUSCLE

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ABSTRACT

A finite element description of blood perfusion has been developed, and is applied to skeletal muscles. Three-dimensional distributions of blood pressures and flows in deforming muscles are calculated. The muscle tissue is considered as a fluid-saturated porous solid. The blood is modeled as a series of five intercommunicating compartmental fluids, representing arterial, arteriolar, capillary, venular and venous blood, that reside in the pores (blood vessels) of the muscle tissue. The blood vessels are modeled as distensible tubes, embedded in the muscle tissue.

A 3-D finite element mesh has been mapped on a reconstructed geometry of a gastrocnemius medialis muscle of the rat. Blood perfused linear elastic muscle material behaviour has been assigned to this mesh. A simulation of blood perfusion, resulting from a constant arterio-venous pressure difference, through the reconstructed muscle has been performed. Calculated blood pressure and flow distributions were within physiological range.

KEYWORDS: *Skeletal muscle – porous media – geometry – perfusion*

INTRODUCTION

In the past decades finite element modeling has become the most popular method for mechanical analysis of morphological structures in biology (Huyghe et al., 1992; Yin, 1985). This is because of its flexibility to combine realistic geometries and complex material behaviour. This study fits into a research program that aims at including micro-structural morphology of biological structures into special purpose finite element models, thereby intensifying the interaction between morphologists and biomechanists.

Blood flow through muscle tissue can be regarded as fluid flow through a porous solid. Thus, porous media theories can be employed to describe the mechanics of deforming, blood perfused muscle tissue. These theories are based on the concept of fluid flow due to pressure gradients through porous solid materials that are saturated by the fluid. As an illustration of this concept, the very early experiment of Darcy and Ritter (1840) clearly shows its essential aspects (Fig. 1). They investigated the relation between fluid flow and pressure gradient for soil specimens. Their results obeyed the following law (Darcy's law):

$$\frac{Q}{A} = K\frac{P_1 - P_2}{L} \qquad (1)$$

where Q is the flow (m^3/s), A is the crossectional area of the soil specimen (m^2), K is the conductivity[1] coefficient of the soil specimen ($m^2/(s\ Pa)$), P is pressure (Pa) and L is the length of the soil specimen (m). The conductivity coefficient of the soil specimen appeared to depend on the pore geometry (size, orientation) and the fluid viscosity. The one-

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dimensional law of Darcy has been generalised to three-dimensional, steady state, Newtonian flow of an incompressible fluid through a saturated porous solid (e.g. (Neuman, 1977)):

$$q = -K \cdot \nabla p \tag{2}$$

where q is the volume specific fluid flow (m/s),∇p the 3-D gradient of the volume averaged pressure (Pa/m), and K the 3-D conductivity tensor $(m^2/(s\ Pa))$, which relates fluid flow (size and direction) to fluid pressure gradient. Flow and pressure are now expressed as volume averaged quantities, representing fluid flow and pressure averaged over all the pores in an arbitrarily bounded local region (averaging volume; Fig. 2). Thus, instead of concentrating on micro-mechanical phenomena within pores (e.g. detailed description of capillary blood flow), a macro-mechanical description of fluid flow between different regions is obtained.

When comparing perfused muscle tissue with Darcy's soil specimen, two major complications concerning the description of the fluid flow arise. Firstly, blood flows through a vascular tree, which consists of a dense bed of arterial, capillary and venous vessels, in each of which blood pressure and flow are totally different. Secondly, the blood is surrounded by distensible vessel walls, which mechanically interfere between blood pressure and hydrostatic pressure in the surrounding tissue. These complications have driven us to the development of an extended description of fluid flow through porous solids, in which 1 the hierarchical architecture of the vasculature, 2 the vessel wall elasticity, and 3 finite deformation of the porous solid are incorporated. The hierarchy is dealt with by subdividing the fluid into five intercommunicating fluid compartments, representing the arterial, arteriolar, capillary, venular and venous blood respectively. Each of these blood compartments is characterized by its own Darcy equation with a specific conductivity tensor, so that compartmental blood pressures and flows can be calculated. Moreover, the intercommunicating flow between the compartments, which corresponds with the physiological definition of regional capillary perfusion $(ml/(s\ 100g))$, is also described by a Darcy-like equation:

$$q_0 = -k_{00} \frac{\partial}{\partial x_0} p \tag{3}$$

where x_0 is a dimensionless parameter that quantifies the hierarchical position in the vascular tree from arteries to veins, and $\frac{\partial}{\partial x_0} p$ represents the hierarchical gradient of the (volume averaged) blood pressure (Pa). The vessel wall elasticity is modeled as a relation between the local compartmental blood volume fraction (local blood volume in a compartment

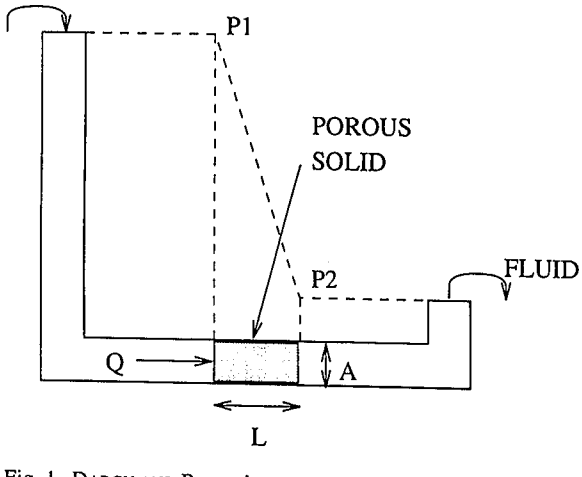

Fig. 1. DARCY AND RITTER'S EXPERIMENTAL SETUP: GLASS TUBE IN WHICH A SOIL SPECIMEN IS FIXED; DUE TO THE PRESSURE DIFFERENCE $P1 - P2$ FLUID FLOWS LEFT TO RIGHT THROUGH THE SPECIMEN.

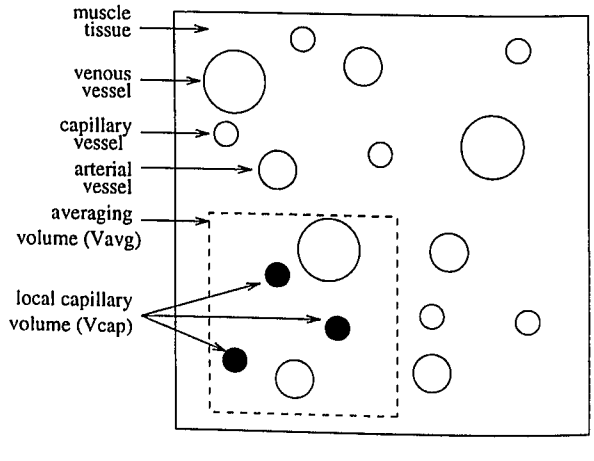

Fig. 2. ILLUSTRATION OF VOLUME AVERAGING PROCEDURE: CROSS-SECTION OF MUSCLE TISSUE IN WHICH SEVERAL TYPES OF BLOOD VESSELS ARE EMBEDDED; AN AVERAGING VOLUME IS DRAWN.

In porous media theory known as permeability.

divided by the local averaging volume, e.g. for the capillary blood: $n_{cap}=V_{cap}/V_{avg}$ (Fig. 2)) and the local transmural pressure difference (local compartmental blood pressure – local extravascular tissue pressure). Thus for each compartment a vessel compliance can be specified that relates the compartmental blood accumulation to the compartmental intra-extra-vascular pressure difference. Intra-extra-vascular exchange of fluid and solutes is not included in the model. Although this exchange is essential from a biological point of view, its influence on the fluid mechanics of blood perfusion is only minimal on the time scales (seconds) that we consider here.

The newly developed model has been implemented in the finite element software package DIANA (DIANA Analysis bv, P.O. Box 113, 2600 AC Delft, The Netherlands).

To perform finite element simulations of perfusion of skeletal muscles, many input data containing geometrical information, material behaviour, vessel wall elasticity and conductivity tensors for the blood perfusion, need to be obtained. In this paper we will focus on geometrical input information, which consists of muscle geometry and geometries of the tendinous structures. This information can be obtained from geometrical reconstruction of a muscle, which has been performed in our group on a rat gastrocnemius medialis muscle. In a companion paper (van Donkelaar et al., 1995) the actual reconstruction procedure is described in more detail. In this paper we illustrate the use of such a geometrical reconstruction in finite element perfusion analysis.

METHODS

A simulation of blood perfusion through rat gastrocnemius medialis muscle was performed. The geometry, which was obtained via precise reconstruction, and its element-division are given in Fig. 3. The mesh consists of 1170 tetrahedrial elements that constitute the muscle tissue, and 130 triangular elements, attached to the muscle surface, representing the aponeuroses. The aponeuroses thicknesses range from circa 0.5 *mm* near the tendon to circa 0.01 *mm* at the other end. The muscle material was modeled linearly elastic, with a Young modulus of 5.0 *K Pa* (Heerkens et al., 1987). The linearly elastic tendinous material of the aponeuroses was modeled much stiffer, with a Young modulus of 1.5 *GPa* (Trestik & Lieber, 1993). The vasculature was modeled by five intercommunicating blood compartments, representing arterial, arteriolar, capillary, venular and venous blood respectively. Blood conduction through the tissue was prescribed for each vascular compartment (*i*) by isotropic spatial conductivity tensors $\boldsymbol{K}_i$ (2), that were constant in the whole geometry. Vascular conductivity k_{00i} (3) was prescribed analogously. The values for $\boldsymbol{K}_i$ and k_{00i} that were used in the simulation are listed in table 1. These values were obtained under the assumption that $\boldsymbol{K}$ is isotropic, and were estimated by (Huyghe, 1986):

$$\boldsymbol{K}=K\boldsymbol{I}\ ;\ K=\frac{d^4}{a_m\ 128\ \mu} \tag{4}$$

where $\boldsymbol{I}$ is the unity tensor, d is the vessel diameter, A_m is the muscle's cross-sectional area and μ the apparent blood viscosity. For the large, single supplying artery and draining vein, $d = 0.5\ mm$, $A_m = 30\ mm^2$ and $\mu = 3.5\ 10^{-3}\ Pa\ s$), by which $K = 5\ 10^{-3}\ mm^2/(Pa\ s)$. Because of the relatively small distances that blood travels through arteriolar, capillary and venular vessels, $K \approx 0$ for these compartments. Thus macroscopic spatial blood flow is only accounted for the arterial and venous compartments. Estimations of the hierarchical conductivity (i.e. vascular conductivity) k_{00i} are made by dividing the expected hierarchical flow (q_0=2.5 $10^{-3}\ s^{-1}$ (Hudlicka et al., 1984)) by the expected compartmental hierarchical pressure gradient $\frac{\partial}{\partial x_0}p$ (Fronek & Zweifach, 1975).

Table 1. Perfusion parameters.

	$K\ (\frac{mm^2}{Pa\ s})$	$k_{00}\ (\frac{1}{Pa\ s})$	$c\ (\frac{1}{Pa})$
arterial	$5\ 10^{-3}$	$5\ 10^{-6}$	1 10–9
arteriolar	$1\ 10^{-6}$	$325\ 10^{-9}$	1 10–9
capillary	$1\ 10^{-6}$	$5\ 10^{-6}$	1 10–9
venular	$1\ 10^{-6}$	$5\ 10^{-6}$	1 10–9
venous	$5\ 10^{-3}$	$5\ 10^{-6}$	1 10–7

Table 1 also contains the values of the vessel compliances c_i, which were also constant in the whole geometry. Suppressing the displacements of the proximal and distal ends of the muscle ensured isometric deformation (Fig. 3). Furthermore 80 *mmHg* arterial blood pressure and 0 *mmHg* venous blood pressure were prescribed in one node at the muscle surface. This node represents the position where the supplying artery and draining vein penetrate into the muscle (Fig. 3). In the analysis the transient development of blood pressures and flows in each compartment in the whole muscle, resulting from the stepwisely applied blood pressure boundary conditions, was calculated. Already after circa 20 *ms* a steady state of blood perfusion was reached, of which some results will be presented.

RESULTS

The main objective of this study is to predict blood pressures and flows in a realistic model of the rat gastrocnemius medialis muscle. The driving force for this blood perfusion is the arterio-venous pressure difference, which is modeled as nodal arterial and venous blood pressure boundary conditions. The resulting blood pressure patterns in the muscle are given in Fig. 4.

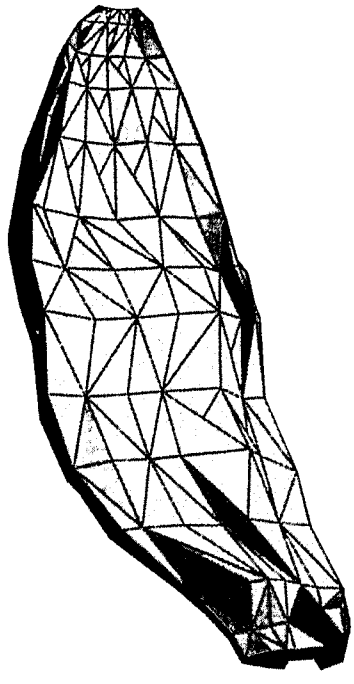

Fig. 3. SHADED VIEW OF THE FINITE ELEMENT MESH OF THE MUSCLE GEOMETRY. Top and bottom edges represent distal and proximal ends, respectively.

Due to the isotropy and large value of the spatial conductivity $\boldsymbol{K}$ of the arterial compartment, the blood is distributed homogeneously and almost without loss of pressure throughout the muscle. The same holds for the venous blood that flows towards the draining vein. (Fig. 4a, c). From the arterial compartment, the blood flows via the arteriolar compartment towards the capillary compartment, between which a significant blood pressure difference exists due to the small hierarchical conductivity k_{00} of the arteriolar compartment (Fig. 4b). The arteriovenous decrease in blood pressure is comparable to experimentally measured blood pressures in resting skeletal muscles (Fronek & Zweifach, 1975). The calculated hierarchical flow q_0 at the capillary level (Fig. 4d) corresponds to the physiological regional capillary perfusion in resting skeletal muscles (Hudlicka et al., 1984).

DISCUSSION

In this paper an illustration of finite element analysis of blood perfusion in resting skeletal muscle has been given. The analysis was applied to a precise reconstruction of a muscle geometry. Roughly estimated input for material behaviour and blood perfusion have been used. Nevertheless reasonable approximations of physiological values of blood perfusion were calculated. Any given muscle geometry can be used in the simulation, which is an important feature of the finite element method.

Another essential feature of the employed finite element model is that contraction and finite deformation of the muscles can be included in the simulations (Vankan et al., 1995). Thus interaction between contraction and perfusion can be studied. Contraction is prescribed as an active stress component in the local fiber direction and the resulting deformation is calculated. The contraction stress is a function of activation, time, local strain and strain rate, dependent of the contraction model that is used. This, however, is beyond the scope of this paper.

Future work should focus on the acquisition of more realistic input parameters, such as muscle material behaviour (e.g. transverse isotropic, non-linearly elastic), more exact anisotropic conductivity ten-

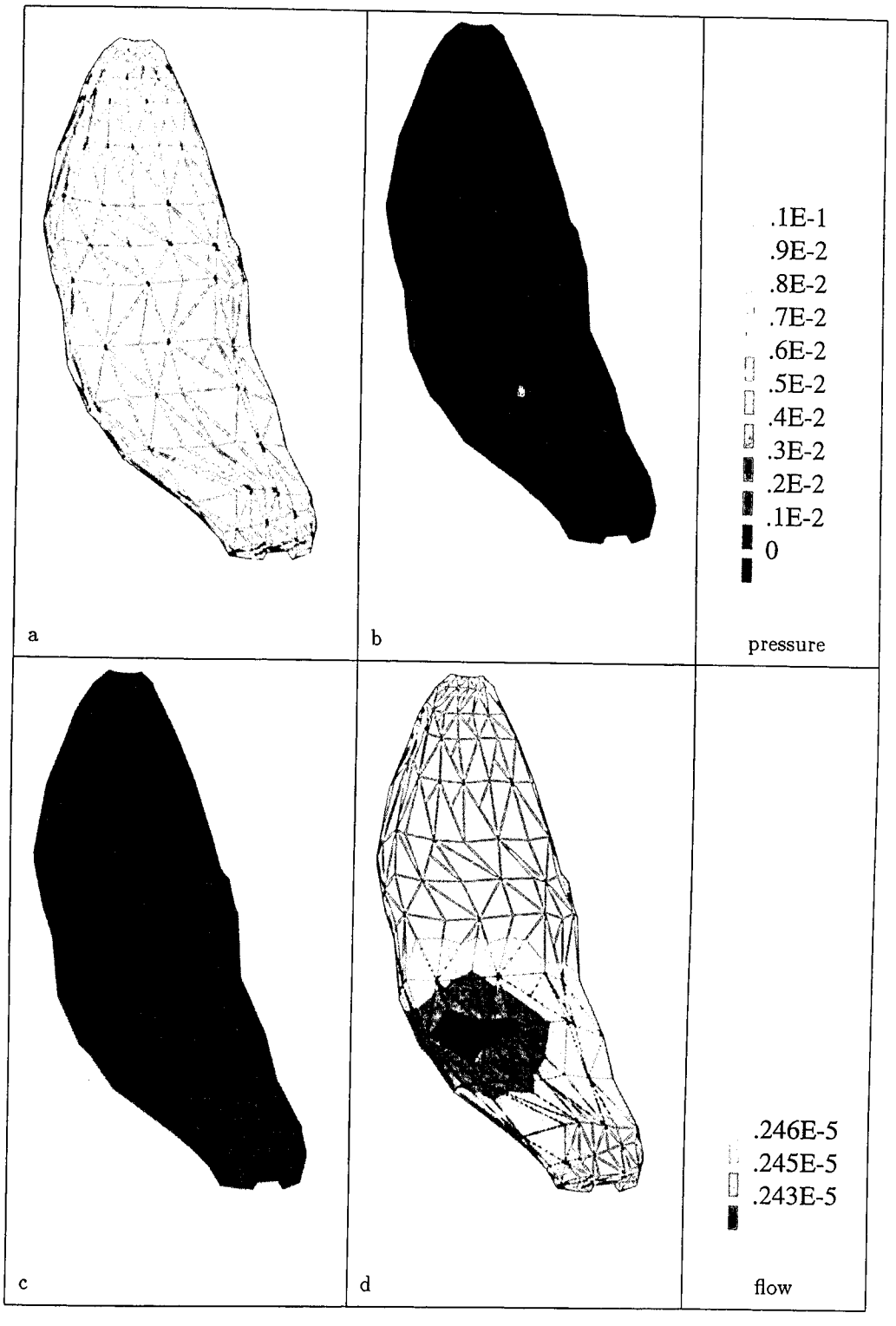

Fig. 4. (A) ARTERIAL, (B) CAPILLARY AND (C) VENOUS BLOOD PRESSURE VALUES (*MPA*), AND (D) CAPILLARY HIERARCHICAL BLOOD FLOW (*1/MS*) (I.E. REGIONAL CAPILLARY PERFUSION).

sors for blood perfusion and more realistic values for compartmental vessel compliances.

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