Národní konference s mezinárodní úcastí

INŽENÝRSKÁ MECHANIKA 2002

13. – 16. 5. 2002, Svratka, Ceská republika

SYNOVIAL FLUID FILTRATION BY ARTICULAR CARTILAGE UNDER CYCLICAL LOADING

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Summary: *The synovial film filtration by articular cartilage in the human ankle joint loaded by a cyclical loading (similar to that in walking) is analyzed. The cartilage matrix is elastic, transversely isotropic and the elastic moduli in compression and tension are depth-dependent. The effect of sliding is not considered and the articular surfaces are assumed congruent. This simplification leads to a one-dimensional (time-dependent) problem. The calculation shows that, within each step, the synovial fluid turns into a thin protecting gel layer that is diluted back again into the fluid. For the normal values of the matrix permeability and of the elastic moduli, the gel-film thickness slowly diminishes with time, after 120 steps being still about one tenth of micrometer thick. The increase of permeability by one order, as with the arthritic joint, however, depletes the protecting gel film after several steps and the surfaces get into the contact.*

Keywords: ankle joint, articular cartilage, cyclical loading, squeeze-film lubrication.

1. Introduction

It has been the focus of many studies to investigate the lubrication properties of articular cartilage in order to understand the normal and pathological behaviours and the reasons for degenerative processes of human synovial joints. To explain marvellous performance of healthy joints, two conflicting conceptions have been proposed, with the fluid transport across the articular surface taken into account. The “weeping” lubrication (McCutchen, 1962) assumes that the synovial film is supplied by the exudation of the interstitial fluid from the compressed cartilage, similarly as of a compressed sponge. On the contrary, the “boosted” lubrication (Walker et al., 1968; Maroudas, 1969) assumes that the solvent component of synovial fluid (SF) flows into the pores of the compressed cartilage. During the squeeze-film action, the concentration of the macromolecular complex of hyaluronic acid and proteins present in the SF increases until a limit value is reached. A fixed synovial gel film is formed then between the surfaces, preventing their intimate contact.

In the last decade, some mechanical models have appeared to model the fluid transport in a loaded synovial joint. These models mostly consider articular cartilage a biphasic mixture of a porous elastic matrix and an ideal interstitial fluid (Mow et al.,

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1980). They revealed an important role of interstitial fluid pressurisation due to loading proposed by McCutchen (1959, 1962). This fluid pressurisation contributes to supporting a significant portion of the total applied load, particularly in the early time response of the material (Ateshian et al., 1994).

Hou et al. (1992) have presented a squeeze-film lubrication model of a rigid impermeable spherical indenter approaching a thin permeable homogeneous cartilage layer supported by a rigid impermeable subchondral bone. A single phase Newtonian fluid for SF and a biphasic model for articular cartilage (porous elastic isotropic and homogeneous cartilage matrix filled with viscous Newtonian interstitial fluid) are used. A perturbation method with two small parameters is applied. Calculations converge for low loads only. After accepting some simplifying assumption, Jin et al. (1992) have succeeded in applying this model to the loads encountered in the human hip joint in standing. The model shows that cartilage porosity depletes the synovial film slightly only.

The biphasic model of cartilage (a porous elastic isotropic and homogeneous matrix and an ideal interstitial fluid) has been applied to an axially symmetric case of the squeeze-film lubrication of the human hip joint in standing (Hlaváček, 1993, 1995, 2000). However, SF is considered a mixture of two incompressible fluids: viscous (hyaluronic acid-protein macromolecular complex) and ideal (water and small solutes). Only the ideal phase passes the interface, thus enabling a synovial gel film description. According to this perturbation model, the filtration by cartilage is intensive, the fluid film gets quickly depleted and a synovial gel layer develops over the greater part of the contact. The gel serves as a boundary lubricant if a sliding motion follows before a fresh SF gets into the contact.

Both the above models corroborate the “boosted” lubrication conception, but the filtration intensity that they predict differs. The reason for this discrepancy seems to lie in the fact that Hou et al. (1992) consider the fluid pressure to be continuous across the articular surface at the moment of a step-load application. They also take only zero-order term in the expansion across the cartilage thickness for the interstitial fluid pressure in the asymptotic solution. Hlaváček (1993) takes into account a small jump in the fluid pressure at the articular surface at the time of step-load application and considers all the second-order terms of the expansions. In fact, the immediate response to a step load of a biphasic cartilage with the incompressible phases is single-phase and incompressible. Not all the boundary conditions at the articular surface valid during a continuous loading can be satisfied at the moment of a step-load application. For example, the total surface traction is continuous, but the fluid pressure is not. For a load increasing continuously from zero, the above jump is absent, but a boundary layer of a large fluid flow remains if the cartilage matrix can expand laterally while being compressed. Another simple example of a jump in the fluid pressure at the surface is the case of an unconfined cubic element of an incompressible single phase elastic material under unidirectional compression. The isotropic Hooke law yields the homogeneous hydrostatic pressure as low as one third of the outer compressive traction.

Macirowski et al. (1994) have measured in vitro the total surface stress and displacement on acetabular cartilage, when step-loaded by an instrumented femoral head prosthesis, together with the surface topography and constitutive properties of the intact cartilage. A simplified finite element model (with shear and lateral strains in the cartilage neglected!) is used to obtain the fluid and solid stress components together with the fluid velocity normal to the articular surface. The latter is obtained from the difference between calculated and measured surface displacements. The results obtained support the "weeping" lubrication concept.

It is apparent that the above papers present quite contradictory results. To close the above literature outline, let us mention the manuscript by Soltz et al. (in submission to Journal of Tribology, ASME). It presents an analysis of the contact of a rippled impermeable indenter against a biphasic cartilage layer. A numerical analysis demonstrates that, under contact creep, the trapped lubricant pool of an ideal fluid depletes within some seconds as a result of an intensive filtration by articular cartilage. A high normal time-decreasing fluid flux across the articular surface shortly after load application indicates the existence of a fluid pressure jump at the surface at the start of creeping, similar as in the model by Hlaváček (1993).

Stress analysis of the contact of isotropic homogeneous cartilage layers shows only partial agreement with the observed failure modes and creep curves. Transversely isotropic homogeneous models that simulate different properties of the matrix in tension and compression, agree more closely with the observed sites of cartilage failure (Donzelli et al., 1999). The latter models predict not only high shear stress at the subchondral-bone interface and separation of the cartilage layer, but are also consistent with lesions observed at the articular surface and provide better curvefits of cartilage early response indentation data (Cohen et al., 1993).

The mechanical properties of the matrix (such as equilibrium tension and compression stiffness, permeability) are also depth-dependent (Akizuki et al., 1986; Schinagl et al., 1997; Setton et al., 1993), in agreement with the matrix microstructure. Especially, the intact superficial zone of the normal AC, with the collagen fibrils oriented along the articular surface, reveals the elastic moduli and permeability considerably different from those of the deep zone. The intact superficial zone of the normal articular cartilage displays a higher tensile stiffness along the surface, lower compressive stiffness and lower permeability perpendicularly to the surface, as compared with the deep zone. Also, a decrease of the tensile and compression stiffness and a significant elevation in permeability, related to the deterioration of the superficial zone in the early human osteoarthritis, have been also reported (Setton et al., 1999).

Some previous studies have suggested that maintenance of a fluid film is significantly aided by the cyclical nature of loading (Medley et al., 1984). The purpose of this paper is to compare the effect of statical and periodical loadings in squeeze-film lubrication of the human ankle joint. In the current application, the cartilage matrix is transversely isotropic, non-homogeneous and a periodic loading, similar to that in walking, is applied. For the sake of simplicity, the articular surfaces are congruent and the effect of surface sliding is not yet considered. This makes the problem one-dimensional only.

2. Formulation of the problem

The human ankle joint can be simply taken as cylindrical (Medley et al., 1983), enabling rotation in the saggital plane only. The joint is represented by two rigid circular cylinders in the inner contact (a cylinder encased in a cylindrical cavity), coated with thin deformable layers (cartilages) of constant thickness $\hat{h}$ under plane strain conditions. The governing equations for squeeze-film lubrication (without any sliding) take the form (Hlaváček, 2000)

$$h_{,t} = -Q_{,x} + 2q, \tag{1}$$

$$\phi_{,t} = -Q\phi_{,x} / h - 2q\phi / h,$$

$$Q = -h^3 p_{,x} / 12\eta(\phi), \quad \eta = \eta_0 (\phi / \phi_0)^w.$$

Here, $t, x, h(x,t), \phi(x,t), p(x,t), q(x,t)$ and $\eta(\phi)$ are the time, coordinate along the synovial film perpendicular to the axis of the cylinders, SF or synovial gel film thickness, volume hyaluronic-acid concentration in the SF, SF film pressure, ideal fluid flux across the articular surface (through the unit surface per second) and apparent SF viscosity, respectively. A comma followed by subscript t and x indicates the respective partial derivative. η depends on ϕ in the form of a power law with the material parameters ϕ_0, η_0 (referring to the non-filtrated state) and w. ϕ varies in the bounds $0 < \phi \le N\phi_0$. $N\phi_0$ stands for the gel-forming hyaluronic-acid concentration that can not be surpassed (Maroudas, 1969).

The fluid flux $q(x,t)$ becomes for the loading applied at $t = 0$ (Hlaváček, 2000)

$$q(x,t) = \beta [p_{,xx}(0^+) / t^{-1/2} + \int_0^t p_{,xx\tau}(x,\tau) \, \mathrm{d}\tau / (t-\tau)^{1/2}], \qquad \beta = \frac{\hat{h}^2}{2\mu} \left(\frac{kE}{\pi} \right)^{1/2}. \tag{2}$$

For congruent surfaces $p_{,xx}$ becomes simple and independent of x in the form

$$p_{,xx}(t) = -3W(t) / 2a^3. \tag{3}$$

μ, E, k are the equilibrium shear modulus, equilibrium confined compression modulus and permeability of the cartilage matrix at the articular surface. $2a$ is the contact length (for the congruency the tibial length), $W(t)$ is the total load per unit width of the joint, applied at $t = 0$ at the contact center and varying with time. In the moment when the gel-forming concentration during the fluid imbibition is reached, the filtration ceases until q given by Eq. (2) changes its sign and the fluid starts to exude. Eq. (2) has been deduced assuming transverse isotropy of the matrix, with E varying quadratically and μ non-varying across $\hat{h}$. Thus, in the present formulation the unknown functions h, ϕ depend on t only ($h_{,x} = \phi_{,x} = 0$). The system of two ordinary differential equations $(1)_{1-2}$ should be solved with the initial conditions

$$h(0^+) = h_0, \quad \phi(0^+) = \phi_0,$$

h_0 being a constant original synovial film thickness, high enough for the solution not to depend on h_0. Eqs. (2-3) have been obtained by an asymptotic method for a thin biphasic layer ($\hat{h}/a \ll 1$). For simplicity, the layer was considered plane (not cylindrical). The method followed the procedure applied for the isotropic and homogeneous case (Hlaváček, 2000). Loading $W(t)$ is taken in the form

$$W(t) = W_0[1 + K\sin(2\pi t / T)]. \tag{4}$$

The sinusoidal part (K and T being the amplitude and period) is superimposed on a step-load W_0. $W(t)$ is a cyclical loading, similar to that in the human ankle in walking.

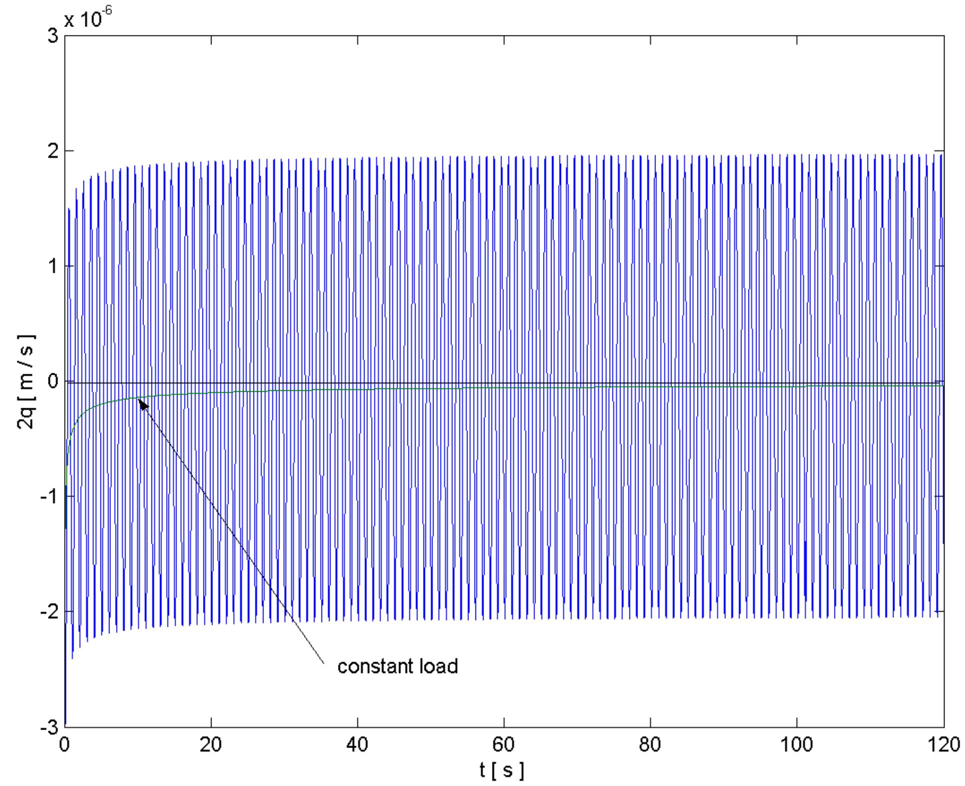

Fig. 1

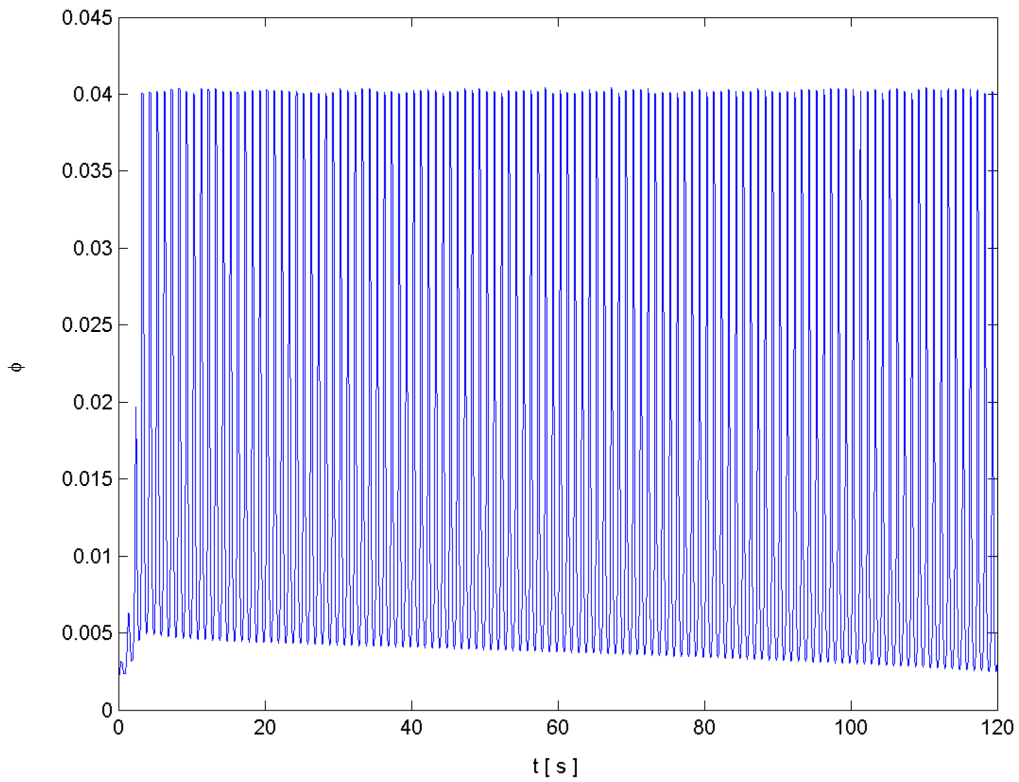

Fig. 2

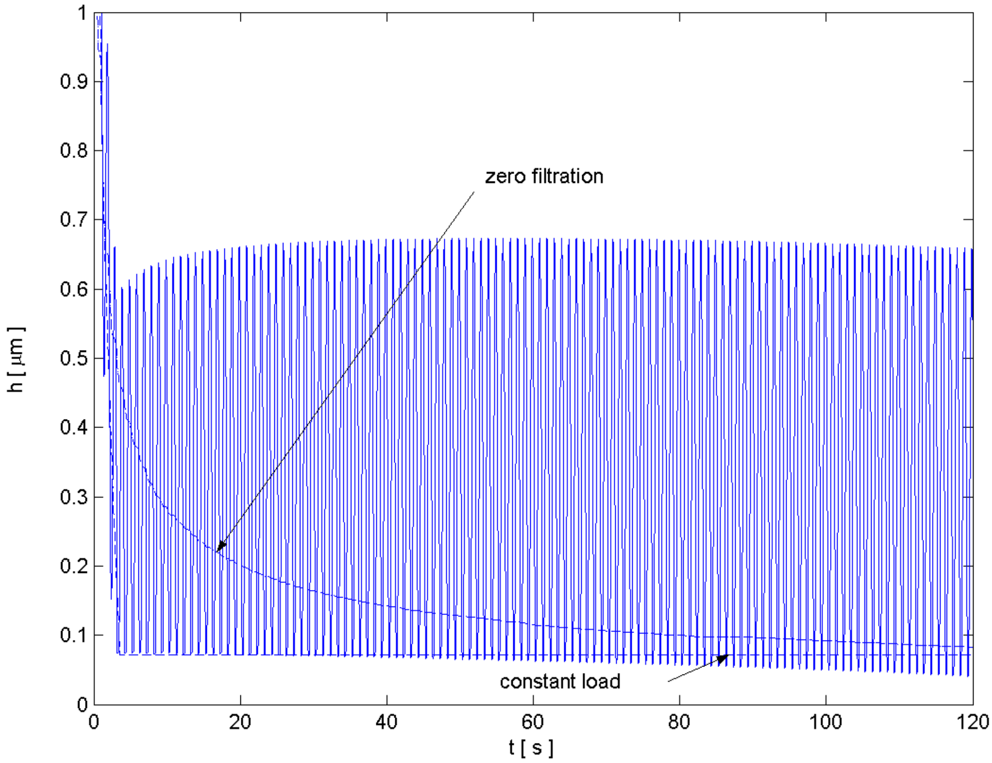
zero filtration
constant load
h [μm]
t [s]

Fig. 3

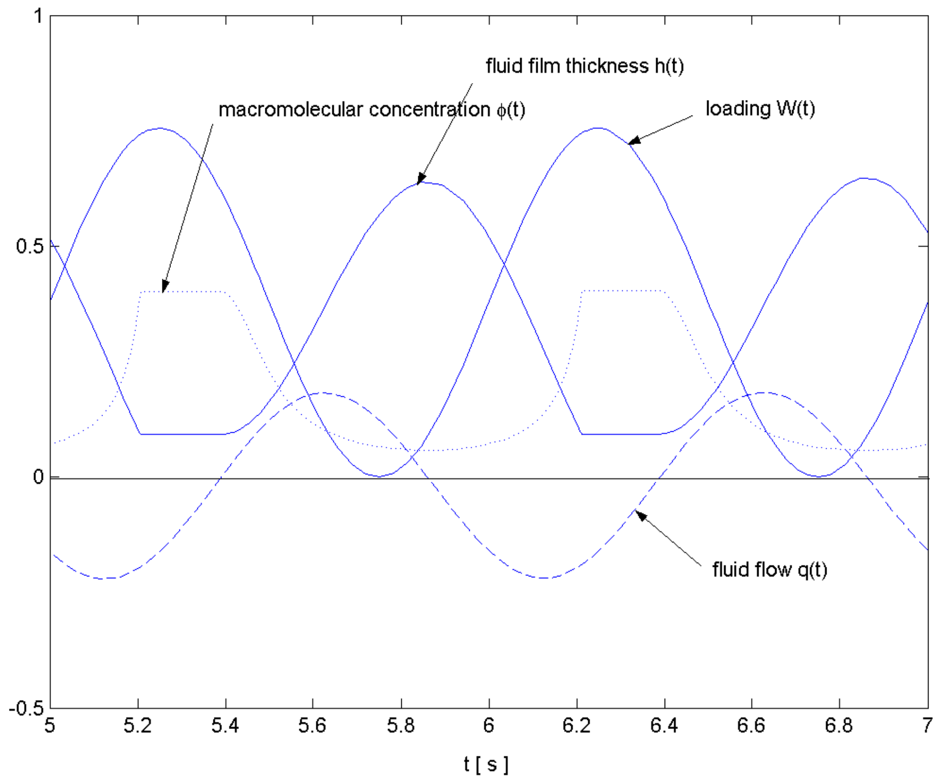
fluid film thickness h(t)
macromolecular concentration φ(t)
loading W(t)
fluid flow q(t)
t [s]

Fig. 4

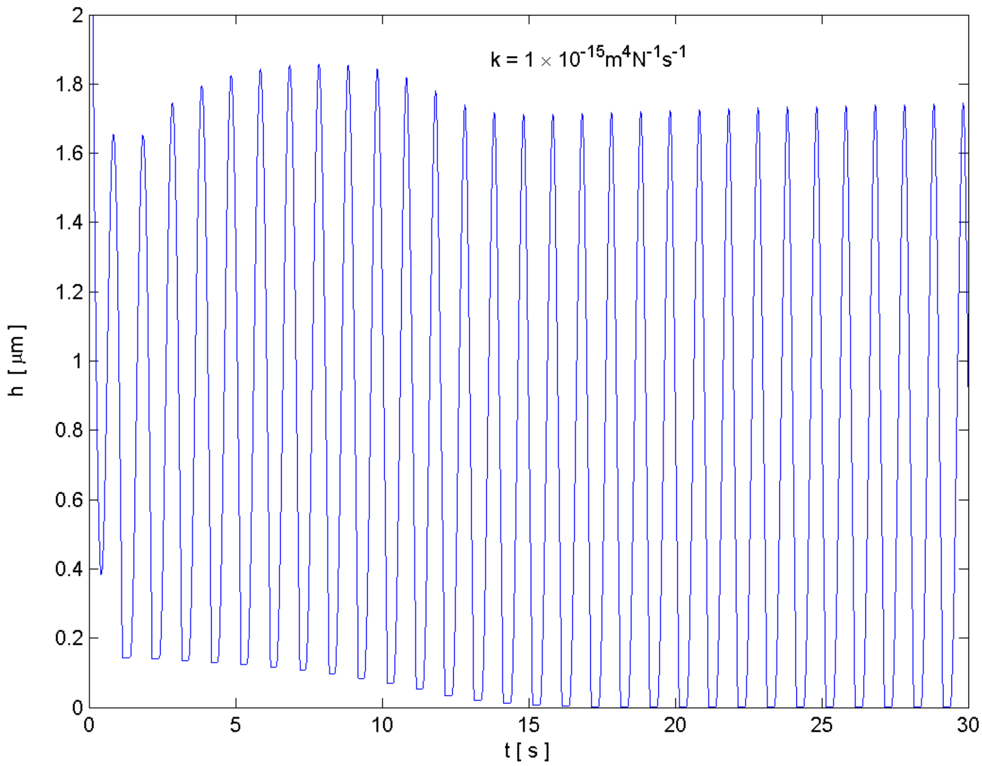

Fig. 5

3. Results

The following set of parameters has been used for the normal ankle joint:

$$\hat{h} = 1.5\text{mm}, \quad a = 1.4\text{cm}, \quad W_0 = 5.4\times10^4\,\text{N/m}, \quad \mu = 0.37\text{MPa},$$

$$E = 0.2\text{MPa}, \quad k = 1\times10^{-16}\,\text{m}^4/\text{Ns}, \quad \eta_0 = 0.004\text{Pas},$$

$$\phi_0 = 0.002, \quad w = 3, \quad N = 20, \quad T = 1\text{s}, \quad K = 1.$$

T is a period of one step. For the tibial width of 2.8cm, W_0 used here corresponds to the total load of 1500N. For $K = 1$ the load oscillates between zero and $2W_0$. The maximum value of $W(t)$ with $K = 1$ is 3000N, a quadruple of the human body weight. Due to the muscle forces, this is an estimated maximum load of the ankle joint in walking. As to the other parameters, they are taken from the literature - see Hlaváček (2000) for the reference.

The above problem has been solved using *MATLAB 6.1*. Fig. 1 shows fluid flow $2q(t)$ across the articular surfaces for the above parameters and 120 steps, calculated from Eqs. (2-4). For a periodic loading with $W(0^+) \neq 0$, $q(t)$ is oscillating and becomes periodic practically after several steps. The case of a constant load, $W(t) = W_0$, is also shown. Fig. 2 shows the change of the hyaluronic-acid concentration ϕ with time. Within each step, the fluid turns into the gel with $\phi = N\phi_0 = 0.04$ and back again into the fluid, due to an intensive filtration. Fig 3 shows the oscillating $h(t)$. The gel-film (minimum) thickness and the maximum value of $h(t)$ within the step interval slowly decrease with the number of the applied steps. In fact, whenever the synovial film becomes fluid-like, the SF flow along and out of the synovial gap restarts. $h(t)$ for zero filtration and for constant loading W_0 are also shown. Fig.4 shows in various scales $h(t), \phi(t), q(t)$ and $W(t)$ for two steps after the fifth step. Variation of $\phi(t)$ is in phase and that of $q(t), h(t)$ are out of phase with $W(t)$. The maxima of $q(t)$ and of $h(t)$ are shifted slightly backwards and forwards, respectively, with respect to the minima of $W(t)$. Fig. 5 gives $h(t)$ for a higher permeability ($k = 1\times10^{-15}\ \mathrm{m^4N^{-1}s^{-1}}$) of the superficial zone. In this last case, the gel layer becomes depleted only after several steps and the articular surfaces may get into the contact, which is an undesirable situation.

4. Conclusion

In the human ankle joint during walking, under the current simplifying assumptions (zero sliding and congruent surfaces), water and low molecular weight solutes flow into (out of) the porous articular cartilage, when loading (deloading) occurs. This flow culminates little earlier than the load. Within each step, the synovial fluid turns into a gel and back into the fluid again.

Arthritic joints show a higher matrix permeability *k*, lower equilibrium compression modulus *E* and lower shear modulus μ in the superficial zone of the articular cartilage than the normal joints (Setton et al., 1993). The fluid transport across the articular surface is proportional to $(kE)^{1/2}/\mu$ for a given SF pressure distribution. It follows that the fluid transport across the articular surface of the ankle joint is higher for arthritic joints. The protecting synovial-gel layer, formed between the articular surfaces at each step, is depleted with the arthritic joints after some steps. However, it is preserved with the normal joints for long, which should be more favorable to keeping the articular surface in good condition.

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The study has been sponsored by the grant No. 103/00/0008 of the Grant Agency of the Czech Republic.