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JOINT CARTILAGE LUBRICATION WITH PHOSPHOLIPID BILAYER

SMAROWANIE CHRZĄSTKI STAWOWEJ Z DWUWARSTWĄ FOSFOLIPIDOWĄ

Key words:

hydrophilic cartilage, lubrication, bi-layer lamellate phospholipids, geometrical and physical data, variations of viscosity of synovial fluid, new effect prognosis

Słowa kluczowe:

chłonna chrząstka, smarowanie, dwuwarstwa fosfolipidowa, dane geometryczne, fizyczne, zmiany lepkości prognozy nowych efektów

Abstract

The surface of an articular cartilage human joint, coated with phospholipid bi-layers or multi-layers, plays an important role in the surface-active phospholipid lubrication, friction, and wear during human limb movement. The biological bi-layer is a thin polar membrane composed of two layers of phospholipids that have a hydrophilic phosphate head (from the outside) and a hydrophobic tail (from the inside) consisting of two fatty acid chains. These membranes are flat sheets that form a continuous barrier around all cells.

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Synovial fluid (SF) in the human joint gap contains glycoprotein, lubricin (protein-glycan 4), and hyaluronidase, i.e. an enzyme that produces hialuron acid and ±10% phospholipids. Because the mechanism of surface articular phospholipid lubrication (SAPL) has been a frequently controversial subject in the past decade, this fact requires showing the hydrodynamic description in the form of a mathematical model of the abovementioned problem and its particular solution. To give a description of this model, it is necessary to recognize the variations of the dynamic viscosity of synovial fluid as a function of parameters depending on the presence of many phospholipid particles. To these parameters belong power (exponent) concentration of hydrogen ions (pH), cartilage wet ability (We), collagen fibre concentration in synovial fluid, and a created electrostatic field on the phospholipid membrane. Based on the Young-Laplace-Kelvin Law, initial achievements presented in scientific papers and our own investigations illustrated in this paper, the decrements, and increments of synovial fluid dynamic viscosities versus pH and wet ability (We) increases, simultaneously taking into account the influence of the intensity of charges in the electrostatic field. Moreover, this study considers the influence of collagen fibre concentration on the dynamic viscosity of synovial fluid. Based on initial considerations performed by virtue of the developed SAPL, it may be stated that the charge increments from low to high values of the electrostatic field is connected with viscosity increases of synovial fluid but only simultaneously with the pH index and cartilage wet ability variations.

JOINT GAP AND PHOSPHOLIPIDS

Synovial fluid in a natural joint gap is limited by the superficial layer and its superficial layer. The lipid bilayers are thin polar membranes made of two layers of lipid molecules. These membranes are flat sheets that form a continuous barrier around all cells. The lipid barrier is the barrier that keeps ions, proteins, and other molecules where they are needed and prevents them from diffusing into areas where they should not be. Biological bilayers are composed of amphiphilic phospholipids that have hydrophilic phosphate heads and a hydrophobic tail consisting of two fatty acid chains. Synovial fluid contains water, ions created after dissociation, collagen fibres, and many various viscous elements, for example, active phospholipids and proteoglycans containing lubricin, which interacts with the hialuron acid [**L. 1–3**]. The lipid layer on the surface of the normal joint cartilage contains mainly Phosphatidylcholine or Phosphatidylserine, which prevents sedimentation effects. Liposomes are composite structures made of phospholipids. The gap geometry and superficial layer is presented **Fig. 1**. (Source: Author's material). **Figure 1** presents the arrangement of forces acting in joint gap filled with synovial fluid between cartilage surfaces, which are limited by the membrane coated by the phospholipid's layer about 1 to 2 nm height. Moreover, the

repulsion force R [N] caused by the negative charge of hydrated phospholipids covering the membrane surfaces is indicated and are the neutralized cations of hydrated sodium occurring in synovial fluid have been observed **[L. 4**]. Mutually cooperating cartilage surfaces have curvilinear cross sections presented in **Fig. 2**. They depended on the shape of joint bonehead and consequently to the type of joint. Various cross-sections of joint surfaces have been considered in relation to the joint, porous-deformations, and roughness changes from 1 μm for infant 2.30 μm for adult to 5.30 μm for osteoarthritis cartilage.

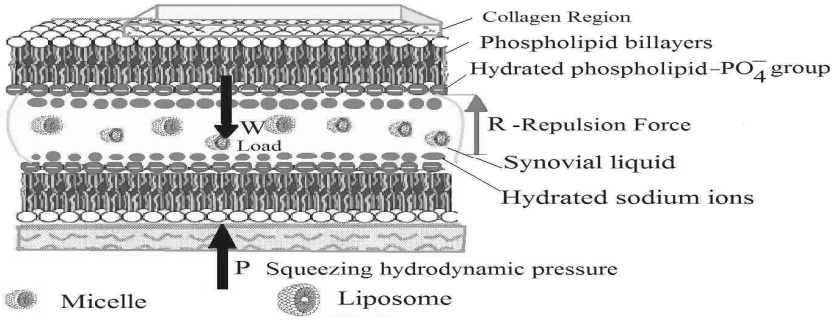

Fig. 1. The joint gap limited by the by the phospholipid bilayer
Rys. 1. Szczelina stawu ograniczona dwuwarstwą fosfolipidów

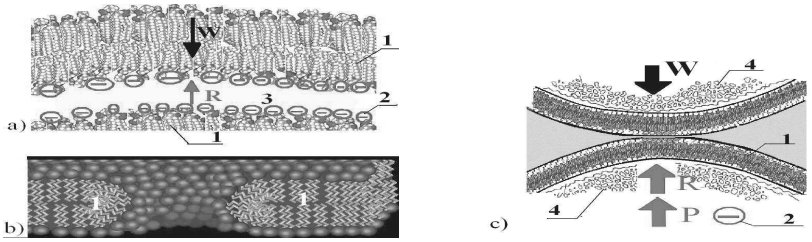

Fig. 2. Phospholipid bilayers on the cartilage surface: a) the curvilinear, b) At the edge of the pore, c) spherical parabolic (knee): 1 – phospholipid layer, 2 – lipid with negative charge, 2 – synovial fluid, W – load, R – repulsion force (own elaboration)
Rys. 2. Dwuwarstwy fosfolipidów na powierzchni chrząstki: a) krzywoliniowej, b) krawędzi parabolicznej pory; c) sferyczno-parabolicznej kolana; 1 – warstwy lipidów, 2 – fosfolipid z ujemnym ładunkiem, 3 – ciecz synowialna, 4 – chrząstka, kolagen, W – obciążenie, R – siła odpychająca, P – siła hydrodynamiczna od wyciskania (źródło: materiał własny autora)

WHAT DETERMINES SYNOVIAL FLUID VISCOSITY?

Traditionally, the pseudoplastic synovial fluid viscosity decreases when the shear rate increases during lubrication flow. However, the synovial fluid viscosity also depends on the concentration of lipids and phospholipids. The

author has not been able to find a publication referring to experimental-laboratory results presenting a direct influence of the percentage of the concentration of phospholipids on the synovial fluid dynamic viscosity. By virtue of the Young-Laplace-Kelvin equation and the author's own research, referring interfacial energy obtained by Z. Pawlak in experimental research [**L. 5–6**], the author of this paper indicates that synovial fluid viscosity directly depends on the concentration of hydrogen ions determined by the Sörensen index pH from 2 to 10, cartilage superficial layer wettability from 40^o to 100^o, and on the concentrations of elastic collagen fibres and lipids and phospholipids. Of secondary importance is the influence of the temperature and electrostatic charge density on the synovial fluid dynamic viscosity, because temperature and electrostatic charge density causes viscosity changes indirectly mutually through interfacial energy, the pH index, and the concentration of collagen or PL particles. In the author's opinion, the direct influence of temperature and electric charge density on the synovial fluid viscosity is negligibly small.

Z. Pawlak **[L. 5**] had experimentally determined the increments of the interfacial energy γ from 1.5 to 3.5 mN/m, for a pH from 1 to 4, and the decrements of the interfacial energy γ from 3.5 to 2.0 mJ/m^2 for pH from 4 to 12. Z. Pawlak comments [**L. 5**] imply from the non-monotone interfacial energy changes, that the amino-group will begin to lose their charge ($-NH_3^+ \rightarrow NH_2$) and the (–POH) group will begin to lose their proton ($-POH \rightarrow -PO^-$), which leaves the surface charged and leading to a decrease in the interfacial energy. This phenomenon implies, in equations (1–2) derived by the author, a new form of transformed Young-Laplace-Kelvin equation, where the interfacial energy γ[mN/m], temperature T [K], wettability We, index pH, the PL surface concentration $s = s_{PL}$ [mol/m^2], synovial fluid viscosity [Pas], and v-synovial fluid velocity [m/s] are mutually connected and explicitly indicated.

$$s(w_e, p_H, T) = \frac{\gamma(w_e, p_H, T)\left(\frac{\gamma_{max}}{\gamma} - 1\right)}{RT} \ln\left[1 + \left(\frac{\sqrt{L_a} - \sqrt{L_b}}{\sqrt{L_k} + 1}\right)^2\right]^{-1}, \quad (1)$$

$$\eta(w_e, p_H, T) = \frac{\gamma_{max} + kA^{-1}T \ln L}{\delta_v \cdot v}, \quad 0 < L \equiv \frac{\left(\sqrt{L_k} + 1\right)^2}{(L_a + 1)(L_b + 1)} < 1, \quad (2)$$

$$L_a \equiv \frac{K_a}{a_H^+}, L_b \equiv \frac{a_H^+}{K_b}, L_k \equiv L_a L_b, \quad (L_a + 1)(L_b + 1) > \left(\sqrt{L_k} + 1\right)^2.$$

Synovial fluid dynamic viscosity varies, usually in interval from 0.003 to 0.5500 Pas. From formulae (1) and (2), the following viscosity increments for $1<pH<3.5\sim4$ and synovial fluid viscosity decrements for $4<p_H<10$ and decre-

ments of viscosity with increments from 70° to 30° of cartilage wettability "We" can be determined. These dependences are illustrated in **Figs. 3** and **4** with an isoelectric point, and are not obtained by virtue of interfacial energy from Z. Pawlak's paper [**L. 5**], but from interfacial energy obtained based on the author's own derivations (1)-(2). Charts of own interfacial energy are not presented. After determining the values for interfacial energy, the author shows the intensive variations of the synovial fluid across the joint gap height presented in **Figs. 5 and 6.**

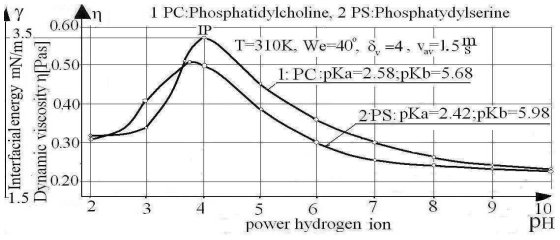

Fig. 3. Dynamic viscosity increments for 0<pH<4 and decrements for 4<pH<10 of synovial fluid versus the concentration of hydrogen ions pH for lipids of type PC, PS, for constant T = 37°C, cartilage wettability We = 40°, average flow velocity 1.5 m/s, and collagen concentration c_c = 500 000 mol/mm^3, IP(γ = 3.5 mJ/m^2, η = 4 Pas). Source: Author

Rys. 3. Wzrosty dla 0<pH<4 oraz spadki dla 4<pH<10 lepkości dynamicznej cieczy synowialnej ze wzrostem stężenia jonów wodorowych dla molekuł lipidowych typu PC i PS dla ustalonej temperatury człowieka 37°C, przy stałej zwilżalności chrząstki We = 40°, stałej koncentracji włókien kolagenowych c_c = 500000 mol/mm^3 oraz średniej stałej wartości prędkości 1,5 m/s, IP (γ = 3,5 mJ/m^2, η = 4 Pas) źródło:badania własne

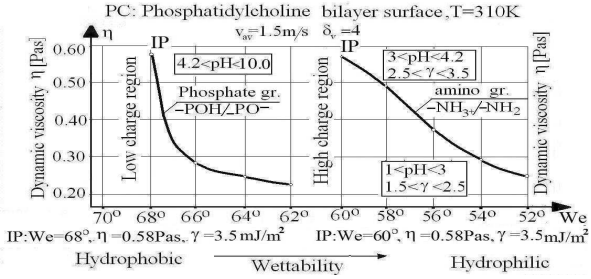

Fig. 4. Dynamic viscosity decrements of synovial fluid versus cartilage wettability for bilayer with phospholipids and amino groups (visible charged surface regions), for a constant temperature of 37°C, constant ranges of pH, and an average human limbs velocity of 1.5 m/s, and a collagen concentration of c_c = 50000 mol/mm^3. Source: Author

Rys. 4. Spadki lepkości dynamicznej cieczy synowialnej ze wzrostem wodochłonności We dla dwuwarstwy fosfolipidów z grupą fosfatową i aminową ze wskazaniem obszarów niskiego i wysokiego ładunku elektrostatycznego od PL w temperaturze 37°C, przy ustalonych zakresach pH, stałej koncentracji włókien kolagenowych c_c = 500 000 mol/mm^3 oraz średniej prędkości 1,5 m/s

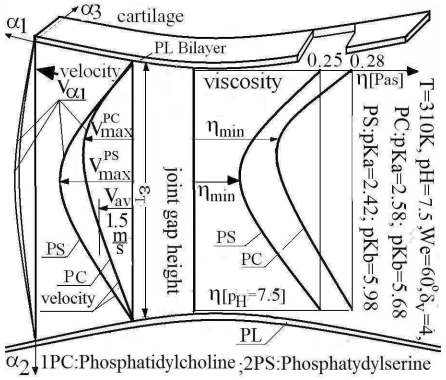

Fig. 5. Comparison of synovial fluid velocity and its dynamic viscosity distributions between two cartilage surfaces in joint gap height direction, for constant value pH = 7.5, for lipids with PC and PS, for a constant temperature of 37°C, cartilage wettability We = 60°, an average flow velocity 1.5 m/s, and collagen concentration c_c = 500 000 mol/ mm^3. Source: Author's own research

Rys. 5. Rozkład prędkości i lepkości dynamicznej cieczy synowialnej między dwoma powierzchniami chrząstek w kierunku wysokości szczeliny stawu dla stałego stężenia jonów pH = 7,5, dla lipidów z PC i PS, dla ustalonej temperatury 37°C, przy stałej zwilżalności chrząstki We = 60°, stałej koncentracji włókien kolagenowych c_c = 500000 mol/mm^3 oraz średniej stałej wartości prędkości przepływu 1,5 m/s. Źródło: wyłącznie badania własne autora

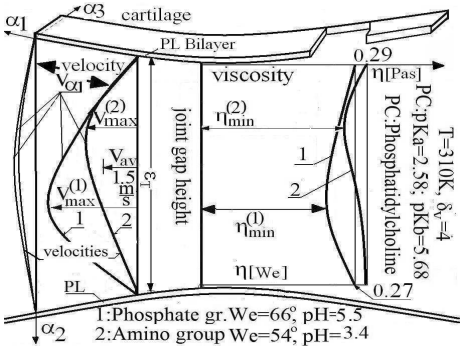

Fig. 6. Comparison of synovial fluid velocity and its dynamic viscosity distributions between two cartilage surfaces in joint gap height direction, for a PL bilayer with phosphate groups when We = 66°, pH = 4.5, and for PL bilayers with amino-groups when We = 54°, pH = 2, at a constant temperature of 37°C, an average flow velocity 1.5 m/s, and collagen concentration c_c = 500 000 mol/ mm^3. Source: Author's own research

Rys. 6. Rozkład prędkości i lepkości dynamicznej cieczy synowialnej z dwuwarstwą fosfolipidów z grupą fosfatową gdy We = 66°, pH = 5,5 z dwuwarstwą fosfolipidów z grupą aminową gdy We = 54°, pH = 3,4, między dwoma powierzchniami chrząstek w kierunku wysokości szczeliny stawu w temperaturze 37°C, dla koncentracji włókien kolagenowych c_c = 500 000 mol/mm^3 oraz prędkości przepływu 1,5m/s. Źródło: wyłącznie badania własne autora

In calculations performed by virtue of equations (1) and (2) we denote the following: gas constant R = 8.3144598 J/(Kmol), Boltzaman constant k = 1.38054 ·(10^{-23} [J/K], A [m^2] cartilage surface coated with the PL molecule, T = 310K, γ[mJ/m^2] interfacial energy, We −wettability of cartilage in interval from 30^o for hydrophilic cartilage to 100^o for hydrophobic cartilage, v-linear velocity of synovial fluid in joint gap (average limb velocity) from 0.25 to 4.0 [m/s], δ_v – a dimensionless coefficient introduced by the author in the range ($2<\delta_v<6$) determining the variations of the concentration of nano-meter long collagen fibres from $\delta_v = 2$ for c_c = 1,000,000 mol/mm^3 to c_c = 100 mol/mm^3 and less for $\delta_v = 6$ in synovial fluid. The calculations assumed average dimensionless values of the ratio $K_a/K_b = L_aL_b$ and the proper dimensionless exponent (power) pK_a, pK_b of K_a, K_b quantities. The K_a, and K_b denoted by Z. Pawlak as acid and base equilibrium constants are, in reality, K_a − the area compression modulus (how much energy is needed to stretch the bi-layer), K_b − bending modulus (how much energy is needed to bend or flex the bi-layer). The a_H − coefficient describing proton activities (hydrogen ion concentration) had been earlier determined by Z. Pawlak. **[L. 4–5]**.

Unfortunately, the results of calculations do not imply which variables independently influence the electrostatic field charge on the increments of the synovial fluid dynamic viscosity without cartilage wettability and without hydrogen ion concentration.

From the formula (1), it follows that the PL surface concentration coefficient $s_{PL}[T,\gamma(pH)]$ decreases when the temperature increases for determined interfacial energy γ in relation to pH. This phenomenon follows from the fact that the high temperature increases the mutually repulsion effect between lipid molecules, thus particle concentration decreases.

From the second formula (2), it follows that the L function attains values in interval (0,1); therefore, (T/A)lnL always has negative values. Thus, temperature increments are decreasing in the numerator of the fraction defining the viscosity, hence viscosity decreases. Analogously, surface A increments denote increments of PL concentration; hence, the negative value (T/A)lnL decrease. This increases the numerator of the fraction describing viscosity, i.e. viscosity increases.

Because velocity increments of synovial flow denote shear rate increments, the denominator increases, and fraction defining the viscosity decreases, i.e. viscosity decreases. This fact confirms the well-known law for pseudo plastic liquids about viscosity decrements with shear rate increments.

The synovial fluid dynamic viscosity varies significantly cross the human joint gap limited by the PL bilayers. The gap height attained average values from 10 to 120 micrometres. Dynamic viscosity increases intensively in the proximity of the bi-layer-phospholipids surface and attained maximum values

in these places. This phenomenon is illustrated in **Figs. 5** and **6.** Based on hydro-mechanical laws, we deduce that the velocity distribution of synovial fluid during the lubrication flow in the joint bearing gap usually has a parabolic profile (shape) with minimal values in the neighbourhood of the superficial layer, i.e. PL-bilayer. From the second formula (2), it follows that, in these places, the dynamic viscosity of synovial fluid attains larger and larger values, because the denominator (fluid velocity) of the fraction presenting the dynamic viscosity has smaller and smaller values. Moreover, in the neighbourhood of the superficial layer, we observe an increase in the concentration of collagen fibres described by the small values of dimensionless coefficient δ_v. From the second formula (2), it follows that the small coefficient δ_v, as the denominator of the fraction presenting the dynamic viscosity, implies large values of the dynamic viscosity of synovial fluid in indicated places. It is visible in **Figs. 5** and **6** that the places of maximal values of synovial fluid velocity profiles coincide with the points where the synovial fluid dynamic viscosity attains the minimal values.

HYPOTHESIS ABOUT THE MEANING OF ELECTROSTATIC CHARGE

The influence of the PL membrane electrostatic field on the functionality of human joint will be considered and examined by virtue of expressions (1-2) as well own and other authors' scientific calculations **[L. 4, 9**]. In the presented example, the electrostatic space charge in synovial fluid attains values in the interval $0<\rho_e<0.04 C/m^2$ for $1<pH<3$, and $-0.10<\rho_e<0\ C/m^2$ for $4<pH<12$. The possible electric current density generated by the electrostatic field in cartilage of a normal patient for pH = 8 has a value of $J = 1.0\ mA/m^2$ [**L. 10**]. If the PL membrane in the joint has small electrical conductivity, $\sigma = 10^{-4} S/m$, from Ohm's and Lorentz's laws, we obtain the electric intensity E and repulsion force R on the surface unit in the cartilage in following form:

$$E = \frac{J}{\sigma} = \frac{1\,mA/m^2}{10^{-4}\,S/m} = \frac{10^{-3}\,A/m^2}{10^{-4}\,S/m} = 10\,\frac{A}{mS} = 10\,\frac{V}{m} = 10\,\frac{N}{C}, \qquad (3)$$

$$R = E\rho_e = 10\,\frac{N}{C}\cdot(-0.1)\,\frac{C}{m^2} = (-1.0)\,\frac{N}{m^2} = -1.0\,\frac{\mu N}{mm^2}.$$

It can be shown that the repulsive force R on a square millimetre of the synovial fluid is very small, one micro-Newton, the R-value is not taken into account during joint lubrication. (Force P>0 for pH>4 and P<0 for pH<4). The author formed the hypothesis that the synovial fluid viscosity variation caused by electrical field and hydrodynamic force P increases. The influence of electric intensity on the fluid dynamic viscosity is described in the following formula [**L. 10–11**]:

$$\eta(\mathrm{pH},\mathrm{We},\mathrm{E}) = \eta_0\left[1+\delta_E(\mathrm{pH},\mathrm{E})\cdot \mathrm{E}^2\right], \tag{4}$$

where δ_E is the coefficient of influence of electric intensity, wettability and pH on the synovial fluid dynamic viscosity. This may be the first time that coefficient δ_E for synovial fluid has been experimentally measured. However, from **Fig. 3,** it is visible that the pH concentration, simultaneously with electrostatic charge, causes approximately 30% viscosity variations. By virtue of this fact and the formula (4), we obtain equality $\delta_E E^2=0.30$. Taking into account value E from formula (3) for pH=8, we obtain $\delta_E=0.003\ m^2/V^2$.

DISCUSSION

In recently publisher papers [**L. 2, 3, 5, 7**] concerning human joint lubrication, we can find numerous papers presenting experimental data referring to the influence of PL concentration, cartilage wettability, and hydrogen concentration pH on the friction coefficient **[L. 4**]. Research indicates that the decrements of friction coefficients are from 1.00 to 0.01, if the PL concentration increases in arbitrary units from 1 to 4. The trial of mathematical methods of the influence of PL concentration on the joint friction coefficient at the molecular level has been undertaken by A. Gadomski et al. [**L. 8**], where, after solving molecular energy equations [**L. 4**], he obtained dependencies determining slow increases of friction coefficient in time, and based on the flexural dissipation law, he was able to show the dependence of viscosity decreasing in time for a constant temperature. It follows that viscosity is inversely proportional to the friction coefficient.

From classical theory of lubrication, it follows that increments of the dynamic viscosity of lubricant implies the increments of load carrying capacity and finally the increment of friction forces. A. Gadomski's research leads to the concept that the increments of phospholipid concentration imply friction coefficient decrements. Z. Pawlak and A. Bojan et al. stipulated that the local large value of dynamic viscosity in synovial fluid could not be in contradiction with the low value of the friction coefficient. Molecular chemistry and publications [**L. 4]** and **[L. 8**], combined with the results of research and the hydrodynamic theory of lubrication and his own experiences inspired the author's further research. According to Amontonson's law, we take the following simple relations between friction force F_R [N], load capacity P[N], and dimensionless friction coefficient μ:

$$F_R = \frac{\eta US}{h_\varepsilon}, \quad P_N = \frac{R_\rho^2\cdot\eta^2}{\varepsilon_T^2\cdot\rho}, \quad F_R = \mu P_N, \quad \mu = \frac{US\rho\varepsilon_T}{\eta R_\rho^2} = \mathrm{Re}\cdot\left(\frac{S}{R_\rho^2}\right) \tag{5}$$

where: η – dynamic viscosity of biological liquid [Pas], U – linear velocity of bio-surface [m/s], S – region of cooperating biosurfaces [m^2], ε_T – average value of total gap height in length units in the range of cooperating surfaces, ρ – density of biological fluid in range from 700 to 1150 kg/m^3, R_ρ – curvature radius of cooperating joint surfaces in m. From this equation (5), it follows that increases of synovial fluid dynamic viscosity implies decreases of the friction coefficient because of the decreases in the value of the Reynold number $Re = U\varepsilon_T\rho/\eta$.

The author's opinion as well the simple general tribological analysis presented by the author in expression (5), indicates that the involved mathematical analysis by A. Gadomski based on molecular-chemical equations without hydromechanical and tribological support are not sufficient to solve the hydrodynamic problem of human joint lubrication with a real phospholipid bilayer. Presently, interesting experimental measurements performed by Z. Pawlak and existing hydrodynamic models of joint lamellar lubrication with numerical solutions for the confirmation experimental results will complete the considered problem.

CONCLUSIONS

This paper presents a new calculation proposal for the human joint lubrication flow of synovial fluid, taking into account its non-Newtonian properties and the amorphous joint gap limited by the quasi-impermeability of the phospholipid membrane. To obtain the real description of the hydrodynamic lubrication of a human joint, this research determined the values of synovial fluid's dynamic viscosities, taking into account the various impurities and various real biological additions and factors, including those at the micro and nano-level, for example, lipids, phospholipids, collagen fibres, hydrogen ions, and other ions after dissociation. Taking into account these factors, the dynamic viscosity of synovial fluid varies, not only in length and width, but also in the direction of gap height. The new results obtained in this research are as follows:

- Joint cartilage wet ability increases (from 70^o to 60^o), i.e. hydrophobic interactions decrease and hydrophilic interactions increase, which imply 50% decrements of the dynamic viscosity of synovial fluid for a constant temperature of 37^oC, a constant concentration of collagen fibres ($\delta_v = 2$), and for hydrogen ion concentration values of $4<pH<12$, and for an average synovial fluid flow velocity of v = 1.5m/s in the presence of the phospholipids bilayers with phosphate groups in the synovial fluid.
- The hydrogen ion concentration increases from $1<pH<3$ and causes the dynamic viscosity of the synovial fluid increments, while the hydrogen ion concentration increases from $4<pH<10$ causes decrements of the

dynamic viscosity of the synovial fluid to about 50%, if the average value of cartilage wettability attained a value of 40^o for an average synovial fluid flow velocity of v = 1.5 m/s in the presence of the phospholipids bilayers with phosphate groups in the synovial fluid.

- The research indicated about 30% variations in the dynamic viscosity of synovial fluid across the joint gap height. Maximal synovial fluid viscosity values are located near the cartilage superficial layer or PL bilayer, because the largest values of lipids concentration, collagen fibres, and electric charges activities are there.
- The effects of the electrostatic charge influence on the synovial fluid dynamic viscosity are caused, if and only if such effects are considered simultaneously with the influences of the hydrogen ion concentrations on the viscosity changes during the joint lubrication.
- After initial prognoses concerning the human joint lubrication performed with a real phospholipid bi-layer and by virtue of the variations of dynamic viscosity of the synovial fluid visualized in 3D in this research, it can be stated that the final determination of the fundamental effects of the hydrodynamic human joint lubrication process in tribological aspects can be obtained after the formulation of a hydrodynamic mathematical model of the human joint lubrication process. Then numerical solutions of load carrying capacities, friction forces, and friction coefficients can be obtained.

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Streszczenie

Dwie kostne powierzchnie trące pokryte chrząstką stawową oddzielone są cieczą synowialną w szczelinie stawu. Hydrodynamiczne smarowanie stawów z udziałem dwuwarstwy fosfolipidów zakłada, że w warstwie wierzchniej chrząstki stawowej istnieją dwie warstwy fosfolipidów o grubości ok. 2 nm w postaci dobrze zorganizowanych molekuł.

Dwuwarstwa lipidowa składa się z dwóch przeciwnie uporządkowanych warstw cząsteczek lipidu z hydrofobowymi końcami węglowodorowymi zwróconymi do środka warstwy oraz polarnymi hydrofilowymi grupami fosforytowymi na zewnątrz. Dwuwarstwa lipidowa jako spontanicznie ukształtowana błona w roztworach wodnych nie przepuszcza związków organicznych i nieorganicznych oraz posiada zdolność gromadzenia równomiernego ładunku elektrycznego po obu stronach, jeśli w roztworze znajdują się jony nieorganiczne . Ciecz synowialna (SF) zalegająca w szczelinie stawu zawiera glikoproteiny, lubrycynę (proteogikan 4), hialuronidazę, czyli enzym produkujący kwas hialuronowy oraz mniej jak 10% fosfolipidów ukształtowanych w postaci liposomów. Na podstawie badań doświadczalnych dotyczących smarowania stawów naturalnych człowieka z uwzględnieniem fosfolipidów liczni autorzy, wykorzystując przeprowadzone eksperymenty natury fizykochemicznej sugerują wzrost sił nośnych oraz zmniejszenie współczynnika tarcia. Nie ma jednak badań podstawowych dotyczących zmian lepkości cieczy synowialnej od jej zwilżalności, pH, prędkości deformacji. Dlatego też w niniejszej pracy autor przedstawił własne badania wstępne odmienne od dotychczasowych. Jako przyczynek do dalszych badań autor wyznacza lepkość mazi stawowej cieczy synowialnej, natomiast nie koncentruje się

na wyznaczaniu energii powierzchniowej celem obliczania poszukiwanych parametrów tribologicznych. Wyznacza się zmiany lepkości cieczy synowialnej w zależności od koncentracji jonów pH i od zwilżalności chrząstki oraz profile rozkładu prędkości i lepkości cieczy synowialnej po grubości szczeliny stawu. Praca niniejsza stanowi wstęp do dalszych badań.