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A model of force transmission in the tibio-femoral contact incorporating fluid and mixtures

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An axisymmetric finite element model is formulated which comprises a rigid spherical indenter, a meniscal ring and an articular cartilage layer, both considered as mixture materials which are interacting with an ideal fluid sub-system.

From parameter studies it is concluded that the application of the mixture theory in comparison with solid modelling only leads to significant effects when the outer surfaces of the components are not sealed. The load distribution appears to change enormously during relaxation of the models. Initially the largest fraction of the load is borne by the fluid in the cavity, while at the end, when the system has reached its final configuration, the meniscal ring bears the major part of the load. Further, the length of the relaxation period appears to depend on the magnitude of the step change of the load. Finally, the curvature of the spherical indenter appears to have significant effects on the loading of the meniscal ring, only immediately after the step changes of load are applied, and these effects disappear as soon as the fluid starts to exude from the models.

1 INTRODUCTION

The present study is intended to provide a model of the tibio-femoral contact complex transmitting loads via the contact between the femur and tibia, both by direct contact of the cartilage-covered articular surfaces and indirectly via the menisci and the synovial fluid. Because of the complex character of the mechanical behaviour of this joint, a stepwise modelling approach is adopted. During every step, parameter studies are performed to investigate the contribution of the relevant components to this behaviour. The first step of the modelling process was reported by Schreppers et al. (1) and consisted of an axisymmetric model comprising the ends of the femur and tibia and a toroid ring in between, representing the meniscus. These models were based on the assumption that load is fully transmitted by direct contact between the cartilage layers and indirect contact via the meniscal ring. All joint components were considered as solids.

The important conclusion of this study was the soft layers at the ends of the femur and tibia exert an important influence on the force transmission in the models studied. For models with soft layers the geometry of the tibial plateau seems to be of negligible importance, and the load borne by the meniscus increases, as compared to corresponding models without soft layers.

In reality both the articular cartilage layers and menisci are not solids but hydrated tissues and the contacts between these components are lubricated by synovial fluid which also has the structure of a hydrated mixture. These tissues comprise free fluid that can move through the solid matrix of the cartilage and the hyaluronic acid–protein complex of the synovia. In the literature the cartilage is often described as a mixture (2–7), while the synovia is considered as a viscous fluid (8). A complex interaction between the cartilage layers and the synovial fluid occurs when the joint is loaded. To account for these effects cartilage and menisci have to be modelled as mixtures.

The present paper considers, with respect to Schreppers et al. (1), the next step in the modelling process of the force transmission in the tibio-femoral contact complex, in which mixtures and fluid are incorporated in an axisymmetric model. A model is defined using finite deformation mixture elements which are coupled to a sub-system consisting of an ideal fluid.

2 DESCRIPTION OF THE MODEL

The model (Fig. 1) is axisymmetric and contains a planar disc, representing the articular cartilage layer, a spherical indenter and, in between, a toroid with a wedge-shaped cross-section, representing the meniscus (9). In the unloaded situation the upper end plane of the articular cartilage layer is in contact with the indenter only at the axis of symmetry. The lower end plane of the cartilage layer is fixed to a rigid foundation. The lower end plane of the meniscal ring rests fully on this layer while the upper surface of the meniscal ring matches the spherical indenter.

Frictionless sliding of the meniscal ring along the articular cartilage layer and the spherical indenter as well as sliding of the articular cartilage layer along the spherical indenter is allowed. Both the articular cartilage layer and the meniscal ring are deformable mixtures of a solid and a fluid. The values for the stiffnesses of the soft layers and the bony components differ enormously and therefore the question arises as to whether it is necessary for the load distribution to take the deformability of the bony components into account. From studies performed by Schreppers (9) the assumption of rigidity of the bony parts in these models does not appear to affect the load distribution. For efficiency reasons this deformability is left out of the model in the present studies: the spherical indenter and the foundation are rigid and impervious. The cavity enclosed by the cartilage layer, the meniscal ring and the rigid sphere is filled with an ideal fluid. For the time being, in the model this cavity is considerably larger than in the...
real knee joint, in order to keep the model initially easier to describe. Fluid flow across the interfaces of the fluid–mixture and mixture–mixture contacts is allowed. The outflow at the outer radius of the meniscal ring and the articular cartilage layer is free while the fluid pressure is assumed to be zero at these places. It is assumed that no fluid layer is present between contacting surfaces. The material behaviour is characterized by a linear coupling between the components \( t_{ij} \) of the second Piola–Kirchhoff stress tensor and the components \( \varepsilon_{ij} \) of the Green–Lagrange strain tensor according to Hooke’s law:

\[
\begin{bmatrix}
\varepsilon_{rr} \\
\varepsilon_{rz} \\
\varepsilon_{zz}
\end{bmatrix} = \frac{1}{E} \begin{bmatrix}
1 & 0 & -v & -v \\
0 & 2(1 + v) & 0 & 0 \\
-v & 0 & 1 & -v
\end{bmatrix} \begin{bmatrix}
t_{rr} \\
t_{rz} \\
t_{zz}
\end{bmatrix}
\]  

(1)

where \( E \) and \( v \) represent the Young modulus and Poisson ratio respectively. Furthermore, the interaction between the solid and the fluid phase is described by Darcy’s law:

\[
\nabla \cdot \mathbf{v}_s - \nabla \cdot (k \nabla \lambda) = 0
\]  

(2)

where \( k \) is the constant permeability coefficient and \( \mathbf{v}_s, \mathbf{v}_f \) and \( \lambda \) represent the gradient operator, the solid phase velocity and the hydrostatic pressure of the fluid in the mixture respectively.

The numerical values for the material parameters are listed in Table 1. Using the equilibrium condition

\[
\nabla \cdot (\mathbf{t} - \lambda \mathbf{I}) = 0
\]  

(3)

together with equations (1) and (2), the deformations in the model and the field of hydrostatic pressure \( \lambda \) can be calculated as functions of time. An axial load of 250 N is applied at time \( t = 0 \) and the response of the model is calculated for the next 25,000 seconds using non-linear finite element techniques. The articular cartilage layer and the meniscal ring are divided into four-node isoparametric mixture elements with linear interpolation functions for the displacements and hydrostatic pressures.

From the performed analysis the load distribution in this model appears to depend strongly on time. In the reference model, presented by Schreppers et al. (1), the load was borne by both direct contact between the femoral and tibial components and contact through the meniscal ring. In the model considered here, a third sub-connection is constituted by the fluid enclosed in the cavity. Figure 2 shows the load distribution over these three sub-connections as a function of time. Initially, just after the load has been applied, the flow in the mixture components has not yet started and about 75 per cent of the total load is borne by the fluid sub-system, while the direct and meniscal contact bear 7 and 18 per cent respectively. As time proceeds fluid can flow out of the model at the circumferential outer surface of the meniscal ring and the articular cartilage layer and the pressure in the fluid sub-system decreases. Because of the outflow, the compression of the total model increases and the fluid sub-system carries a continually decreasing part of the total load. Finally, after 10,000 seconds the hydrostatic pressure is approximately zero all over the model and the load is borne only by direct contact (16 per cent) and the meniscal ring (84 per cent). Flow across the upper surface of the cartilage layer only occurs in the regions where it is contacting the fluid sub-system or the meniscal ring. The outflow ceases in the zone where it is contacting the impervious spherical indenter.
The meniscal ring has a dual function in this model. Initially, just after the load is applied the enclosure of the fluid cavity by the meniscal ring results in pressure being built up in the cavity, while finally, when fluid pressure approximates zero, the meniscal ring carries the larger part of the total load.

3 PARAMETER STUDIES

In comparison with the models presented in the first step of the modelling process (1), in this work an additional function for the meniscal ring with respect to the building up of pressure in the cavity is found. To gain more insight into this function, parameter studies have been performed with respect to the interface conditions and the load that is applied to the model respectively. In the following the consequences of the transition from the solid material models to the mixture material models are studied systematically.

4 INTERFACE CONDITIONS

Firstly, the interface conditions with respect to the fluid flow are considered here. Higginson and Norman (6) questioned the necessity of taking into account the cartilage components in the tibio-femoral joint as mixtures because of their very low permeability. Both the meniscus and cartilage layer comprise a dense network of fibres at their outer surfaces and it might be speculated that this layer hampers the fluid flow across these surfaces.

Five models are considered which differ from each other with respect to the boundary conditions for the hydrostatic pressure. Two basic parameters are considered. Firstly, there is the presence or absence of fluid in the joint cavity. The other concerns the fluid flow. As well as models in which fluid flow across the outer surfaces of components is allowed, models with sealed mixture components and models with solid components are also considered. In model A (Fig. 3) the meniscal ring and articular cartilage layers consist of solid material while the cavity is empty. Model B is similar to model A, with the cavity being filled with an ideal fluid. Models C and D are similar to models A and B respectively, except that the meniscal ring and articular cartilage layers are mixtures which are sealed at their outer surfaces. Finally, model E is the model presented in the previous section, which is identical to model D apart from the fact that the outer surfaces are not sealed. The version of model C without sealed surfaces is not taken into account here. The reason for this is that it is considered to be a non-realistic model as no pressure is built up in the cavity although the space is enclosed. The constitutive behaviour of all models is the same as that of the model described in the previous section on the understanding that models A and B have zero permeability.

In Fig. 4a and b the undeformed and deformed element meshes for models A and B are shown. The axial compression of both models is clearly visible and larger for model A than for model B. The contact area between the rigid sphere and the tibial component is larger for model A than for model B. These effects can
be deduced from the larger axial stiffness of model B as a result of the load-bearing capacity of the fluid in the cavity. The volume of the cavity remains unchanged in model B while it is reduced with increasing load for model A.

Figure 5a and b represents the principal stresses related to the Cauchy stress tensor for models A and B in the integration points. Compressive stresses in the \( r_z \) plane are indicated by dashed squares while the tensile stresses are represented by squares built up from solid lines. Compressive stresses in the circumferential direction are shown by crosses made up from dashed lines whereas tensile stresses in the circumferential direction are shown by crosses made up from solid lines. In both models the stresses in the meniscal ring are mainly directed circumferentially. In model B (cavity filled) at the inner side of the meniscal ring the larger compressive stresses in the \( r_z \) plane are mainly radially directed, while for model A they are more axially oriented at this place.

Figures 6 and 7 show the axial compression and the fraction of the total load that is transmitted by the meniscal ring respectively versus time for all five models. Because fluid flow is absent in models A and B, their behaviour is constant with time. From Fig. 6 the axial stiffness for the models with a fluid-filled cavity (B, D and E) appears to be larger than for the models with an empty cavity (A and C), because the fluid bears part of the load. For the former models the loading of the meniscal ring is smaller. Initially, the curves of model E are very close to the curves of model D, but as time proceeds more and more fluid is squeezed out and at the end the hydrostatic pressure equals zero everywhere, yielding the same conditions as in model A.

5 LOADINGS

In the parameter studies described by Schreppers et al. (1), the joint load ranged from 0 to 1000 N. The load distribution appeared to depend on this load. In this work all analyses were done for a step change of the load from 0 to 250 N. The influence of the magnitude of the step is still unknown. Two possible effects are proposed beforehand. The first effect concerns an increasing total compression of the model for larger loads. Thus more fluid has to be squeezed out of the model, resulting in a larger time to elapse until no further changes occur as a result of the step change of the load. In the following this period is called the relaxation period. The other effect implies an increase of the initial
pressure in the cavity for larger loads. The resulting fluid velocities will increase so that the relaxation period is expected to be shorter.

Taking these contradictory effects into account, it cannot be predicted whether the relaxation period will increase or decrease for larger loads. Therefore, the effect of the magnitude of the load is investigated by performing an additional analysis with model E for a loading step of 500 N.

In Fig. 8 the pressure in the cavity is given as a function of time. Figure 9 shows the fraction of the load transmitted by the meniscal ring versus time for both loadings. From both figures it can be seen that the relaxation period for the loading of 500 N is approximately half of the relaxation period for the loading of 250 N. After relaxation the fraction \( q \) is smaller for the higher loadings than for the lower loadings, which is consistent with the results of the solid models described by Schreppers et al. (1).

6 SURFACE GEOMETRY

Both for the reference model and for the reduced model, presented by Schreppers et al. (1) and Schreppers (9) respectively, the curvature of the contact surface(s) appeared to have only minor effects on the load distribution in the models when the contact surfaces are covered by soft layers. This is an important characteristic for the mechanical behaviour of the tibio-femoral contact complex. Therefore, we want to know whether it also applies when mixture materials are used and the cavity is filled with fluid. Starting from model E, two models with different curvatures of the rigid spherical indenter are defined. These models are indicated by F.
Fig. 5 Principal stresses in integration points. See text for further explanation.

Fig. 6 Axial compression of the models A to E versus time for a load step of 250 N at \( t = 0 \) s.
and G and their radius is 20 and 60 mm respectively. Thus, in model G the wedge of the meniscal ring and the cavity are smaller while for model F they are larger. To both models a step change of the load from 0 to 250 N is applied. After the load is applied, some time will elapse until only the solid materials bear the load. Then the fraction of the load borne by the meniscal ring can be expected to be approximately the same for the models E, F and G. Whether this will be the case for the initial response, when the fluid bears part of the load, is not clear beforehand.

In Fig. 10 the parts of the load borne by the meniscal ring versus time for the models E, F and G are shown as results of the performed numerical analyses. From this figure the initial loadings of the meniscal ring appear to differ for the three models up to 10 per cent.

The relaxation period for the model G, with the larger radius, is smaller, while the relaxation period for the model F with the smaller spherical radius is larger in comparison with model E.

7 CONCLUSIONS AND DISCUSSION

From these analyses the following conclusions can be drawn:

1. The application of the mixture theory on the model of the tibio-femoral contact complex only leads to significant effects if the outer surfaces of the components are not sealed.
2. If these surfaces are not sealed the load distribution depends on the load history in such a way that the
3. The fluid-filled cavity carries a large part (up to 75 per cent) of the total load applied on the model and this fraction decreases to zero when the fluid is being squeezed out of the model.

4. The length of the relaxation period appears to depend on the magnitude of the step change of the load in such a way that it decreases for increasing load steps.

5. Finally, in the presented models, variation of the curvature of the spherical indenter appears to have significant effects on the loading of the meniscal ring, just after the step changes of load are applied. These effects disappear when the fluid is leaving the models.

Because of the strong abstraction of the actual models, care must be taken when translating these characteristics to the real knee joint. The simplifications in the models pertain to the two-dimensional nature of the geometry, the chosen values of material properties, the fluid–solid interaction and the absence of meniscal attachments to the tibial plateau. The load values and loading rates were primarily chosen to demonstrate the capabilities of the present models. Any attempt to relate them to some characteristics of identified human movement, for example walking or running, would be highly speculative. However, it is supposed that the performed analyses properly describe the effects of basic parameters in the tibio-femoral contact complex.

To achieve a more realistic description, two main aspects should be considered in future steps of the mod-
elling process of the force transmission in the tibio-femoral contact complex. The first aspect concerns the interaction of fluid and matrix in the hydrated tissues. Because electrolytes are dissolved in the fluid phase of the articular cartilage and the proteoglycan aggregates of the solid matrix are ionized, local concentrations of electrical loadings are created by forcing the fluid to move out of the model. Therefore, the time that elapses until the load distribution is stationary will probably be smaller in reality.

The other aspect concerns the possibilities of fluid outflow from the cavity when load is applied. In reality, fluid may escape more easily because the meniscus will not fully enclose the cavity, as is the case in the model. Although quantitative data about the resistance against the fluid outflow are not yet available, these possibilities should be the subject of research in further steps of the modelling process. The authors of the present paper are convinced that this can only be done in an appropriate way if these flows are considered as unknown quantities in the model and are implicitly calculated for the applied loadings. Therefore, the viscosity of the fluid has to be taken into account.

Finally, a possible way to arrive at a complete model of the tibio-femoral contact complex needs to be described. Firstly, attention has to be focused on the interaction of mixture components with a Newtonian fluid. Then squeeze film effects in both the direct and indirect contact area have to be investigated. This aspect could cause the relaxation period of the model to decrease considerably. Next, the application of more complex loading patterns, such as harmonic axial loads, can be performed. When a three-dimensional formulation is applied, bending of the knee may also be simulated. Finally, more detailed descriptions of the material and geometry of the real joint can be implemented. In this stage, for example, physical non-linear material behaviour and effects resulting from moving electrical loadings in the articular cartilage can be taken into account.

REFERENCES