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A rheologic model of the menisci in static and dynamic load transmission

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The important load-carrying function of the meniscus in the knee joint is frequently being emphasized in modern orthopaedic literature. However, not much descriptive data on this function are known.

To evaluate its mechanical performance, extensive experiments with fresh cadaveric pig knee joints were carried out. Different types of loading

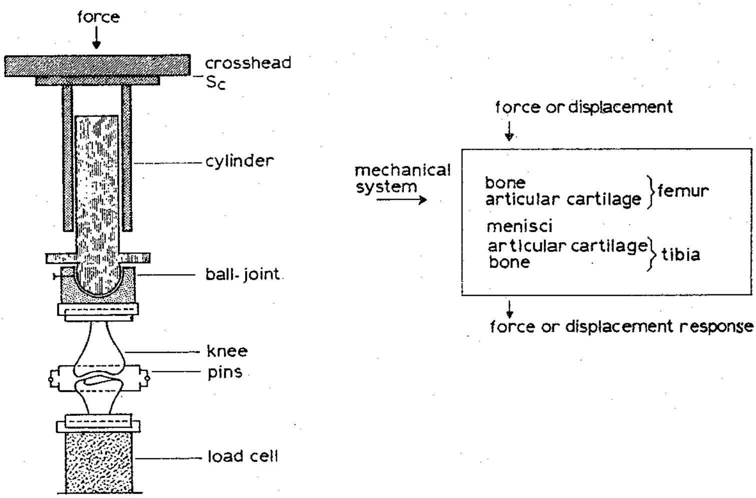

Figure 1. Schematic drawing of the experimental setting. On left: The knee joint is fixed in a specially designed apparatus that fits into an Instron testing machine. To the pins, extensometers are connected to measure the displacement response upon loading. On right: Description of the joint as a mechanical system.

functions were applied to the knee joint in extension, with intact knees and after meniscectomies. Slow and fast loading rates (ramp loading and step loading), repeated step loading and impact loading were used as experimental loadings. The transient deformation response of the joint was recorded and analyzed. Figure 1 shows a schematic drawing of the experimental setting. Figure 2 shows an example of the deformations between the pins resulting from step loading, for a knee with menisci and after a meniscectomy. Furthermore, load-carrying contact areas in the knees, with and without menisci, were measured as a function of the loading, using a special roentgen technique. Parts of these experiments have been reported previously (Jaspers et al., 1978). Figure 3 shows the load-carrying area as a function of the loading for two different knees, with and without menisci.

In order to be able to describe the characteristic parameters of the system and roughly evaluate their influence on its mechanical behavior, so as

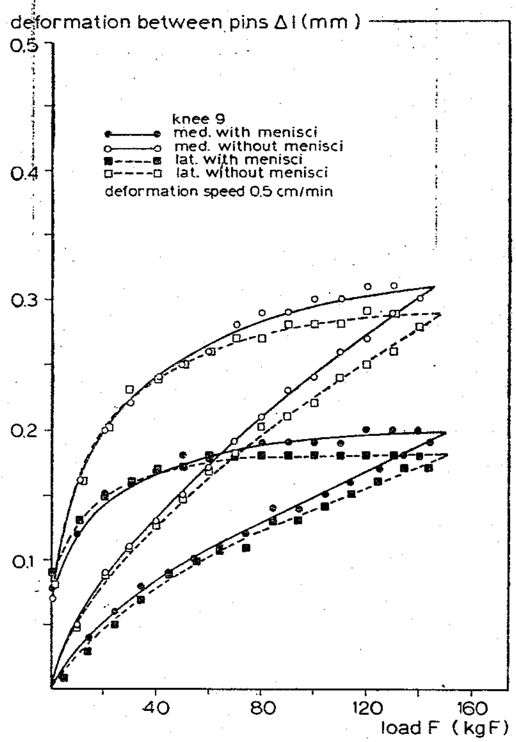

Figure 2. Deformations between the pins as a function of the applied loading, at the lateral and the medial sides, for a knee with menisci and after meniscectomy. Curves for increasing load (starting at $\Delta 1 = 0$) and decreasing load are shown.

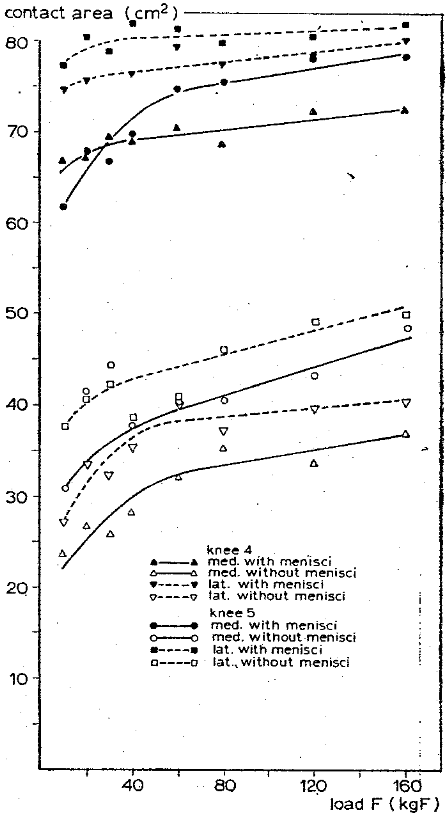

Figure 3. The load carrying contact areas in the medial and lateral parts of two knee joints, with menisci and after meniscectomy, as a function of the applied load, as measured with a special roentgen technique. It is evident that meniscectomy drastically reduces the contact area, and hence increases the average stress normal to the contact surfaces.

to quantify the mechanical function of the menisci, a mathematical (rheologic) model was developed.

This model was partly based on the results of geometrical measurements and a (limited) amount of data in the literature on the mechanical properties of the joint components and further developed by adjusting the model characteristics to the experimental results.

It was assumed that the state of deformation was uni-axial; hence the mid-saggital plane was one of symmetry. It was found that the mechanical influence of the menisci can be separated into two effects : a non-linear, but time-independent effect related to the circumferential stretching and a non-

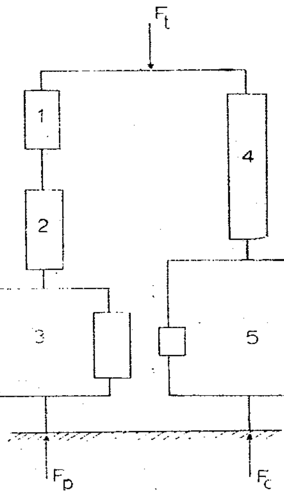

Figure 4a. The rheologic model of the knee in extension. The force-transmission mechanism across the central part of the knee (direct contact between tibia and femur) and the peripheral part (menisci) can be represented as parallel impendances. The meniscus effect can be represented by two impendances in series, as discussed in the text.

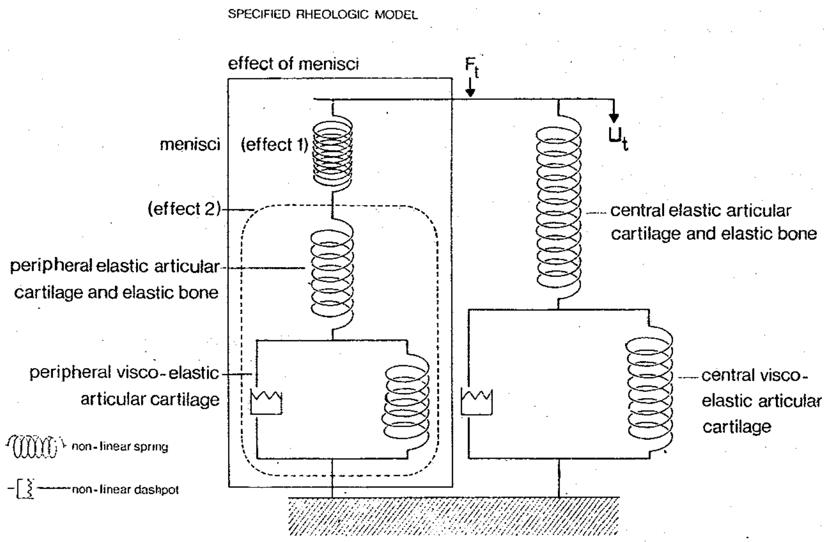

Figure 4b. Definition of element displacements and forces ($F_t = F_p + F_c$).

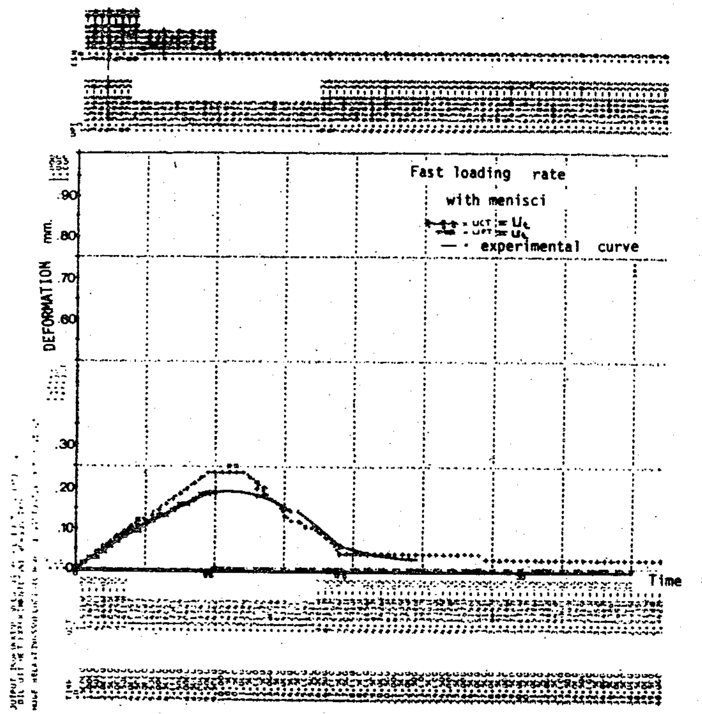

Figure 5a. A comparison of experimental (drawn lines) and model (dotted lines) results: displacements as a function of the time in fast loading (step loading): intact knee.

linear visco-elastic effect related to the load-carrying area and the material properties of the articular cartilage.

The complete rheologic model, consisting of non-linear springs and visco-elastic elements, is shown in Figure 4. The material properties of the elements are described by formulas in which certain parameters appear but the physical significance in some cases is still uncertain.

In the following formulas the element displacements are denoted by u (mm), the forces by F (kgf):

Element 1: $\quad u = hF^s;$

Element 2: $\quad u = z_3\left(\frac{F}{A_p}\right)^{z_4};$

Element 3: $\quad F = \frac{A_p}{l_p \cos^2\alpha}\left\{\left(b_{p_1} + b_{p_2}\frac{F\cos\alpha}{A_p}\right)\frac{du}{dt} + \left(k_{p_1} + k_{p_2}\frac{F\cos\alpha}{A_p}\right)u\right\};$

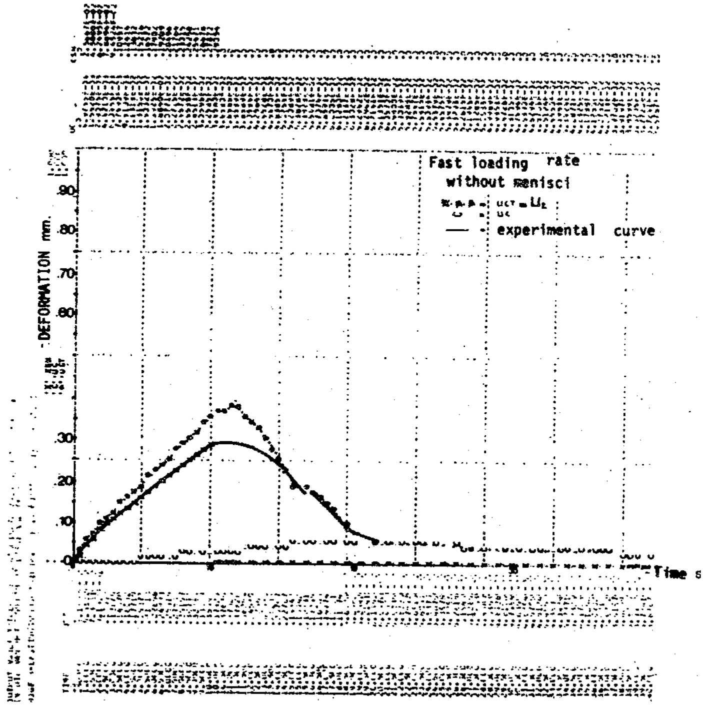

Figure 5b. A comparison of experimental (drawn lines) and model (dotted lines) results: displacements as a function of the time in fast loading (step loading): after meniscectomy. Meniscectomy in the model is simulated by removing Element 1 as shown in Figure 4.

Element 4: $$u = z_1\left(\frac{F}{A_c}\right)^{z_2};$$

Element 5: $$F = \frac{A_c}{l_c}\left\{\left(b_{c_1}+b_{c_2}\frac{F}{A_c}\right)\frac{du}{dt}+\frac{A_c}{l_c}\left(k_{c_1}+k_{c_2}\frac{F}{A_c}\right)u\right\},$$

where: A_c and A_p (mm^2) are the central and peripheral contact areas, respectively;

l_c and l_p (mm) the central and peripheral articular cartilage thickness (total of tibial and femoral parts);

α is half of the meniscus apex angle, h, s, z_1 through z_4, b_{c_1}, b_{c_2}, b_{p_1}, b_{p_2}, k_{c_1}, k_{c_2}, k_{p_1}, and k_{p_2} are parameters of which the physical significance is uncertain.

It was shown that the value of A_p (the peripheral (or menisci) contact area) is independent of the load. A_c is a function of the central force $(A_c = A_c(F_c))$. The general equations for the model can be expressed as:

$$f_1 = f_2 u_t + f_3 \frac{du_t}{dt} \quad \text{and} \quad f_4 = f_5 u_t + f_6 \frac{du_t}{dt}$$

where

$$f_1 = f_1\left(F_p, \frac{dF_p}{dt}\right), \quad f_2 = f_2(F_p), \quad f_3 = f_3(F_p),$$

$$f_4 = f_4\left(F_c, \frac{dF_c}{dt}\right), \quad f_5 = f_5(F_c), \quad \text{and} \quad f_6 = f_6(F_c)$$

It can be shown from the contact area measurements, that by approximation

$$F_c = \lambda F_t \quad \text{and} \quad F_p = 1 - \lambda F_t$$

where λ depends on the direction cosines of the (linearized) area-load curves, according to

$$\lambda = \text{dir. cosine without meniscus/dir. cosine with meniscus}$$

From geometrical measurements it was found :

$$l_c \simeq 3\text{ mm}; \quad l_p \simeq 2\text{ mm}; \quad \alpha \simeq 15°$$

Using literature data (e.g., Krause, 1976) it was found :

$$h \simeq 0.006 \quad \text{and} \quad s \simeq 0.6$$

Values for the other parameters were estimated on the basis of a part of the experimental results, as shown in Table 1. In using this model to simulate all other experiments (for which a computer program was set up), good agreement was found for all knees, in a qualitative sense. An example is shown in Figure 5. The model can be regarded as a good basis for further research, especially where human circumstances are concerned. Since for human menisci the apex angle is much less, it can be anticipated that in this case the effect of the circumferential stretching (Element 1) is not very pronounced.

Table 1. Estimates for the parameter values, as based on the experimental results (when forces are expressed in kgf, displacements in mm)

Parameter	Value	Parameter	Value
z_1	1.10	b_{p_1}	3.78
z_2	0.71	b_{p_2}	451.16
z_3	0.57	k_{c_1}	0.18
z_4	0.69	k_{c_2}	1.33
b_{c_1}	7.03	k_{p_1}	0.097
b_{c_2}	690.78	k_{p_2}	1.02

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