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Nonlinear Dynamic Behavior of the Human Knee Joint—Part I: Postmortem Frequency Domain Analyses

Characteristic results of postmortem experiments on five knee-joint specimens are reported. The experiments were performed to investigate the applicability of a local linearization technique that would make it possible to describe the dynamic behavior of the joint in terms of transfer functions. The results indicate that the stiffness of the bracing wires, attached to muscle tendons to create a static equilibrium position, can be accounted for when determining the stiffness of the joint. Besides the static equilibrium configuration, the magnitude of the dynamic load and the type of dynamic load applied to the joint can be shown to have their influence. As the influence of the dynamic load is significant, it has to be concluded that in essence the knee joint has to be regarded as a nonlinear system, making application of a Local Linearization Technique questionable. However, when the magnitude of the dynamic load is included as an additional measurement parameter, an indication can be obtained about the behavior of the joint and the degree of nonlinearity.

1 Introduction

Although in many human activities the loading of the human knee joint is of a dynamic nature, little fundamental research has been done on the dynamic behavior of this joint (Hefzy and Grood, 1988). The presence of highly incongruent load bearing joint surfaces together with menisci, ligaments, capsule, and muscles make the knee one of the most complex joints in the musculoskeletal system. Apart from their importance for maintenance of joint stability, the different joint elements may also influence the transmission of dynamic loads. The work reported in this paper is part of a long-term research project aiming at the development of an experimentally validated model of the dynamic behavior of the human knee joint, with parameters that can be interpreted in terms of the properties and functions of the joint elements and interactions between the elements.

After reviewing literature on this subject one cannot but endorse the statement made by Hefzy and Grood (1988) that "... the lack of experimental data needed in the determination of model system parameters and in the validation of the model, is the main reason why mathematical knee joint modelling has not progressed further."

Because of the limited amount of experimental data and validated models for the dynamic behavior of the human knee joint an attempt has been made to gain more insight into the behavior of the joint by means of experiments. It was decided

to start an explorative experimental investigation into the mechanical characteristics of the human knee joint under dynamic loading. It should be recognized that in general the dynamic behavior of the joint must be described by means of 1) relationships between loads exerted on the joint (loads due to muscular activity, and loads due to inertial and gravitational effects), 2) the 3-dimensional position and orientation of the tibia relative to the femur and 3) deformations of and stress distributions in the joint elements, which are related by their individual constitutive behavior. All these relationships will in general be time-dependent and nonlinearly coupled, where it can be assumed that the knee joint behaves as a nonlinear system with nonlinear and time-dependent stiffness and damping characteristics. The nonlinearities can be of a geometrical as well as of a physical nature due to the large movements of the femur with respect to the tibia and the nonlinear material characteristics of the joint elements.

In an explorative experimental investigation the dynamic behavior described above is much too complicated to deal with, due to a lack of experimental techniques to quantify all parameters and quantities governing the relationships mentioned. Therefore some restrictions had to be made. For a start it was assumed that a Local Linearization Technique (LLT) might yield an appropriate experimental procedure. First results obtained from this procedure have been reported in an earlier paper (Jans et al., 1988). The LLT basically consists of two steps. The first step involves the creation of a stable, static equilibrium configuration of the joint by means of a static load. To include one important effect of muscular activity this

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static load can also be used to generate a compressive preload, which may have considerable effect on the transmission of dynamic forces through the joint. The second step in the LLT is the application of dynamic loads such that only relatively small changes in the joint configuration will occur. It is assumed that these changes are small enough to obtain a dynamic behavior that corresponds to the behavior of a linear system with constant mass, damping and stiffness characteristics. These characteristics may depend on the static equilibrium configuration and the static load exerted, but it is essential that they are constant for a certain range of the magnitude of the applied dynamic loads. This range must be determined experimentally.

This approach is attractive because an experimental tool, the so-called Transfer Function Analysis (TFA), is available for the analysis of linear systems, yielding a full description of their dynamic behavior (Bendat and Piersol, 1980).

It must be realized that this approach may fail for two reasons. First it is possible that the behavior of the joint is essentially nonlinear such that nonlinearities arising from nonlinear damping or nonlinear static load-displacement characteristics cannot be linearized. Beforehand this is not expected to occur as the friction in the knee joint is found to be very small (Radin and Paul, 1972) and because a LLT has been applied by Crowninshield et al. (1976), Moffat et al. (1969) and Pope et al. (1976), to describe joint behavior without giving indications for possible failure.

Second, generally available measurement techniques provide a threshold for the minimal magnitude of dynamic loads to be applied as below this threshold the signal-to-noise ratio considerably decreases. This finite measurement accuracy may make it impossible to apply the small dynamic loads required to allow for the local linearization. Because of the lack of knowledge of the dynamic behavior of the joint, at this juncture it was not possible to judge whether these factors would play a disturbing role. Consequently it was decided to start with the experimental procedure described above bearing in mind the possible reasons for failure mentioned.

2 Experimental Procedure

An extensive description of the experimental setup, used in the experiments and shown in Fig. 1, has been given already (Jans et al., 1988). Figure 1 shows the specimen mounted in this setup.

An electro-mechanical shaker (1) was used to apply a small dynamic load to the tibia. The accelerations of the tibia were measured by means of accelerometers mounted on a cylinder in which the tibia is cast. The dynamic load was chosen to be a zero-mean random signal generated by means of a computerized data-acquisition system (Jans et al., 1988). Its variance σ_f could not be set lower than approximately 1 N^2 because of a poor signal-to-noise ratio in the measured accelerations. Under the small dynamic load the displacements of the tibia were small too: they were hardly observable by eye and therefore were not likely to result in geometrical nonlinearities (which are to be avoided when using the LLT).

Following a standard procedure the experiments started with the preparation of the joint specimen the day before the actual measurements were done. This preparation took about 4 hours. The description given below is valid for all specimen used in the experiments described. All knee joint specimens were excised from macroscopically intact human cadavera, freshly frozen at -80°C . Each specimen had an overall length of 0.4 m in extension, with an equal length of the tibia and the femur of 0.2 m. The day prior to the experiments the specimen was thawed in approximately 4 hours in water of 20°C . The preparation started with removal of the skin and subcutis. Subsequently the heads of the muscles were removed, such that the relevant muscle tendons were mobilised but left intact (the

Fig. 1 The upper picture shows the electro-mechanical shaker (1), suspended in the hook (2) of a travelling crane, connected to the tibia by means of a flexible connection. This connection is shown in detail in the lower picture: a = force transducer, b = flexible string, c = flexible hose, d = magnet, e = head of the electro-mechanical shaker, R = stainless steel cylinder into which the distal part of the tibia is fixed, g = gravity.

tendon of the rectus femoris, biceps femoris, gracilis, sartorius, and semitendinosus muscles). During preparation care was taken to keep the joint capsule and extra-capsular ligaments intact. To be able to exert the static load on the muscle tendons by means of bracing wires a minimum length of approximately 0.1 m of the tendons was required. After preparation the specimen was stored overnight at 5°C in Ringer's solution and used in the experiments the day after.

3 Experimental Results

Measurements were carried out on five knee joint specimens (denoted KNEE1 through KNEE5), varying the static equilibrium position (controlled by means of the static preload), the magnitude of the static preload, the magnitude of the dynamic load and the direction in which the dynamic load was applied. Because of autolysis the total measurement time was limited to two days. Therefore not each of the influencing parameters mentioned above could be analyzed for each knee joint specimen. In the sequel a number of results are presented which are considered of prime importance regarding 1) the question whether the LLT is a valid procedure and 2) to find essential characteristics of the knee joint. Results for experiments on KNEE1, KNEE3, KNEE4, and KNEE5 are described in the subsequent sections, focusing on these two key topics. Therefore a discussion on variability between different knee joint specimens is omitted as this was not the prime objective of the experiments carried out. KNEE2 was used for experiments in which selected joint elements were damaged. An extensive discussion on the influence of these operations on the dynamic behavior of the joint is omitted here because this is described in more detail in a separate paper (Dortmans et al., 1991).

The static forces acting on the muscle tendons are denoted by F_r , F_b , and F_a representing the force on the rectus femoris muscle, biceps femoris muscle, and the pes anserinus, respectively. The flexion angle of the joint is denoted by ϕ (Jans et al., 1988). The results are given in terms of transfer-functions H_{uf} between the applied load f_e and the displacement of the tibia u ($u = X, Y$, or Z) in a specific direction relative to the static equilibrium configuration: $H_{uf}(f) = u(f)/f_e(f)$, where f is the frequency in Hz (Jans et al., 1988). As shown in a previous paper (Jans et al., 1988) the frequency interval of

Fig. 2 Influence of the diameter of the bracing wires upon the transfer function $H_x(f)$
 0.75 mm — KNEE3: $\sigma_f = 20$; $\phi = 60$ deg; $F_b = 131$; $F_r = 200$; $F_a = 133$
 1.00 mm --- KNEE3: $\sigma_f = 20$; $\phi = 60$ deg; $F_b = 131$; $F_r = 200$; $F_a = 133$

Fig. 3 Influence of the diameter of the bracing wires upon the transfer function $H_y(f)$
 0.75 mm — KNEE3: $\sigma_f = 6$; $\phi = 30$ deg; $F_b = 120$; $F_r = 210$; $F_a = 133$
 1.00 mm --- KNEE3: $\sigma_f = 6$; $\phi = 30$ deg; $F_b = 120$; $F_r = 210$; $F_a = 133$

interest is from 5 to 50 Hz. In this frequency range two distinct resonance frequencies of approximately 20 and 28 Hz could be detected which correspond to two vibration modes of the tibia denoted mode I and mode II, respectively.

3.1 Influence of the Bracing Wires. Figures 2 and 3 give results for experiments on KNEE3 in which the diameter of the bracing wires was changed from 1.0 to 0.75 mm. From Fig. 2 follows that vibration mode II is hardly affected by the stiffness of the bracing wires, whereas Fig. 3 indicates the opposite for mode I. For mode I the modal stiffness K (the combined stiffness of the joint and the bracing wires) was determined for both cases by means of a curve-fit procedure (Mergey, 1980). The modal stiffness for a diameter of 0.75 and 1.0 mm, $K_{0.75}$ and $K_{1.0}$, was calculated as 14500 N/m and 10470 N/m, respectively. Because the transversal stiffness of the bracing wires is negligible, only their longitudinal stiffness is relevant. If this stiffness would dominate the intrinsic stiffness of the joint, $K_{0.75}$ and $K_{1.0}$ would be proportional to the square of the diameter of the wires. From the values given above, it is concluded that this does not apply. Hence it is concluded that the stiffness of the bracing wires cannot be neglected, but on the other hand is such that the contribution of the joint to the modal stiffness can be determined. In a numerical model this influence can be incorporated and therefore no modifications were introduced.

3.2 Influence of the Static Equilibrium Position. Various experiments have been conducted to study the influence of the static equilibrium configuration. Figures 4 and 5 show some typical results for KNEE3. It is observed that changes in the static equilibrium position result in measurable changes in the transfer functions. For both mode I and mode II, increase of

Fig. 4 Influence of the flexion angle of the joint upon the transfer function $H_x(f)$
 KNEE3: $\sigma_f = 20$; $\phi = 20$ deg; $F_b = 131$; $F_r = 200$; $F_a = 133$
 KNEE3: $\sigma_f = 20$; $\phi = 30$ deg; $F_b = 131$; $F_r = 200$; $F_a = 133$
 KNEE3: $\sigma_f = 20$; $\phi = 60$ deg; $F_b = 131$; $F_r = 200$; $F_a = 133$

Fig. 5 Influence of the flexion angle of the joint upon the transfer function $H_y(f)$
 KNEE3: $\sigma_f = 20$; $\phi = 20$ deg; $F_b = 131$; $F_r = 200$; $F_a = 133$
 KNEE3: $\sigma_f = 20$; $\phi = 30$ deg; $F_b = 131$; $F_r = 200$; $F_a = 133$
 KNEE3: $\sigma_f = 20$; $\phi = 60$ deg; $F_b = 131$; $F_r = 200$; $F_a = 133$

the flexion angle leads to a decrease in the resonance frequency, indicating a reduction in the resistance of the joint against applied forces.

3.3 Influence of the Static Preload. Another parameter influencing the dynamic behavior of the knee joint is the magnitude of the static preload exerted by the bracing wires. Figure 6 shows the transfer function $H_y(f)$ for KNEE1 for three levels of the static preload. Obviously an increase of the forces acting on the muscle tendons results in an increase of the resistance of the joint such that the resonance frequency for mode I increases.

3.4 Influence of the Dynamic Load. A key factor to judge the applicability of the transfer functions given in the preceding sections is to which extent the magnitude of the dynamic load influences the results. In a previous paper (Jans et al., 1988) results of preliminary experiments were reported, suggesting that the magnitude of the dynamic load had only a minor influence. More extensive research has been devoted to this particular point in the experiments on KNEE3 through KNEE5. Figures 7 and 8 give the transfer functions for two experiments on KNEE4. Although the coherence function, corresponding to these transfer functions, always attained values between 0.95 and 1.0, these results show that the influence of the magnitude of the dynamic load cannot always be neglected. Such findings make use of a LLT questionable. To verify that the results are not caused by the non-deterministic nature of the loading pattern (random excitation), additional experiments were carried out on KNEE5 with $\phi = 30$ deg using sinusoidal excitation with varying amplitude (Fig. 9). The results from these experiments confirm the results discussed before (both in a qualitative and a quantitative sense). It may therefore be concluded that the nonlinearities observed are not bound to a particular excitation signal.

Fig. 6 Influence of the static preload upon the transfer function $H_y(f)$
 KNEE1: $\sigma_f = 20$; $\phi = 20$ deg; $F_b = 40$; $F_r = 90$; $F_a = 60$
 KNEE1: $\sigma_f = 2$; $\phi = 20$ deg; $F_b = 84$; $F_r = 210$; $F_a = 90$
 KNEE1: $\sigma_f = 2$; $\phi = 20$ deg; $F_b = 224$; $F_r = 396$; $F_a = 244$

Fig. 7 Influence of the dynamic load applied to the tibia upon the transfer function $H_y(f)$
 KNEE4: $\sigma_f = 7.5$; $\phi = 30$ deg; $F_b = 118$; $F_r = 240$; $F_a = 97$
 KNEE4: $\sigma_f = 15$; $\phi = 30$ deg; $F_b = 118$; $F_r = 240$; $F_a = 97$
 KNEE4: $\sigma_f = 25$; $\phi = 30$ deg; $F_b = 118$; $F_r = 240$; $F_a = 97$

Fig. 8 Influence of the dynamic load applied to the tibia upon the transfer function $H_y(f)$
 KNEE4: $\sigma_f = 7.5$; $\phi = 30$ deg; $F_b = 118$; $F_r = 240$; $F_a = 97$
 KNEE4: $\sigma_f = 15$; $\phi = 30$ deg; $F_b = 118$; $F_r = 240$; $F_a = 97$
 KNEE4: $\sigma_f = 25$; $\phi = 30$ deg; $F_b = 118$; $F_r = 240$; $F_a = 97$

4 Discussion of the Results

In this paper a number of aspects of the dynamic behavior of the knee joint have been discussed. Using the LLT proposed, the dynamic behavior of the joint can be quantified, although it turns out that linearization of the behavior of the joint is not allowed for the range of the dynamic loads applied to the tibia. The transfer functions for various load levels show a marked shift, indicating a decrease of the stiffness of the joint with increasing load level. The value of the results is that they essentially give a good insight in the behavior of the joint. Also an order of magnitude for stiffness and damping parameters can be obtained.

An important observation is that the results for various knee

Fig. 9 The transfer function $H_y(f)$ obtained for experiments with sinusoidal loading on KNEE5 with $\phi = 30$ deg.
 $\bullet$ $|f_e| = 1.8$ N; $+$ $|f_e| = 2.4$ N; $*$ $|f_e| = 3.0$ N

joint specimens may well be compared. Also the results for random and sinusoidal loading agree fairly well. It must be kept in mind however that the obtained results depend on the magnitude of the applied dynamic load.

Due to the use of Transfer Function Analysis (which is an averaging process) and random excitation, the transfer functions must be seen as to give an "average" behavior of the joint for the loads applied. The nonlinearities of the behavior of the joint are translated into a kind of effective stiffness and damping. As the coherence function for all measurements was almost unity, the transfer functions seem to give an acceptable description for the behavior of the joint, despite the nonlinearities involved. This may partly be due to the use of random excitation which is considered as a rather "gentle" excitation (compared to impulse- or step-like excitation). A description of the behavior of the joint using the transfer functions obtained may therefore be considered. As was done in the work of van Heck (1984), who encountered similar difficulties analysing the strongly nonlinear dynamic behavior of slideways, in this case the stiffness and damping characteristics of the joint can be calculated for various levels of the dynamic load applied. Such an approach may yield valuable information on the degree of non-linearity. It is noticed however that in this case no direct insight is gained, as the nonlinearities may partly be smoothed due to the use of spectral analysis and random excitation. This disadvantage may to a certain extent be avoided by application of a time-domain analysis technique. Using measured time domain signals, the best fitting linear system can be determined to obtain stiffness and damping values for various levels of the dynamic load applied. Evidently such an approach may result in discrepancies between the measured and best fitting signals, which can only be resolved by use of a non-linear dynamic model of the knee joint. Such a model is not available, however, and must be focussed on in future research. Therefore the use of a best fitting linear model is considered in the end as an inevitable alternative, that nevertheless may provide valuable information for the development of a nonlinear dynamic model of the knee joint.

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