40. EAN
40. MEZINÁRODNÍ KONFERENCE EXPERIMENTÁLNÍ ANALÝZY NAPĚTÍ
40th INTERNATIONAL CONFERENCE EXPERIMENTAL STRESS ANALYSIS
3. – 6. VI. 2002, PRAHA/PRAGUE, CZECH REPUBLIC

Contact stress in the human hip – mathematical model compared to direct measurement

Kontaktné tlaky v bedrovom kĺbe – porovnanie matematického modelu a experimentálnych meraní

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Keywords: hip, biomechanics, contact stress

Introduction

It was suggested that too high contact stress in the human hip is an important factor accelerating development of coxarthrosis. In the previous studies *in vivo* loading of the hip (contact stresses) were determined by mathematical modelling [1] or by direct measurement [2]. The contact stress distributions in the hip were measured directly by an instrumented prothesis [2,3]. This approach is complicated and offers no benefit to patient. Therefore mathematical models may be adequate for estimating global contact stress distribution. Verification of the mathematical models consists of comparison of the results of calculations with results of measurements.

Methods

Contact stresses from an instrumented endoprothesis

Carlson et al. [3] developed radio telemetry device which can monitor the magnitude and distribution of the stress generated between the cartilage of the acetabulum and the surface of the hip prosthesis. The stress distribution was detected by an array of 14 sensors integrated in the surface of the ball of prothesis.

The pressure-measuring endoprothesis was implanted in 73 years old patient who had sustained a displaced fracture of the femoral neck [2]. Body weight of the patient was 68 kilograms. Data were collected throughout period of recovery and rehabilitation.

Mathematical model

In our work a model developed recently was used to determine distribution of the contact stress in the human hip. The model is described in detail elsewhere [4].

Inputs into our model are: movement of the body of patients and the hip joint resultant force transmitted between the acetabulum and the femur. The hip joint force resultant was determined by the measurement with the implanted instrumented prothesis [5]. These measurements were carried out by professor Bergmann in Germany in four patients [6]. Patients performed several activities and together with the measurement of the force the video motion analysis was performed to determine kinematics of the segments of the body.

Results and discussion

In the table 1 values of the maximum peak contact stress in the human hip obtained from direct measurement and from the mathematical model are shown. The highest contact stress was observed during climbing stairs in the direct measurement as well as in the mathematical modelling.

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Activity	Measurement	Mathematical model
	p_{max} [Mpa]	p_{max} [MPa]
Walking	1.74	3.8
Standing	1.69	3.1
Up stairs	2.10	10.2
Standing up	1.30	9.2

Table 1: *Maximum peak contact stress p_{max} in routine activities.*

Differences between values in the table 1 could be caused by different factors. In the mathematical model several simplifications were introduced. We did not take into account acetabular notch and anteversion of the acetabulum which decrease size of the weight-bearing area and the maximum contact stress as well.

The main difference between could be caused by the inconsistency of compared data. Data of the hip joint reaction force used for calculation of the contact stress distribution were taken from the measurement with implanted instrumented prothesis in four patients [5]. Every loading cycle was performed several times by each patient. These single trials were then averaged. So our calculation of the contact stress represents an average value of peak contact stress. Instead of our results Hodge [2] reports the maximum value of contact stresses that were measured.

Input into used mathematical model is geometry of the hip besides other parameters. In calculation of the contact stress distribution the geometry of operated person was unknown and therefore standard geometry was used. It was shown that the geometry of the hip significantly influences the contact stress distribution [7]. Therefore different geometry in the calculated hip than in the operated one cause the different stress distribution.

In the operated hip the artificial metal ball replaces the femoral head covered by a cartilage. Due to different configuration of the operated hip the contact stresses could be different as well. The head of the prothesis probably does not match the acetabulum close. It could be especially important by extreme movements during which the hip joint reaction force is inclined from the frontal plane. So the high differences between computed values and measured values in standing form a chair and walking upstairs could be explained.

Conclusion

The importance of the standard examination techniques in the biomechanics is indicated. Standard processes in evaluation of the human movement should be introduced. As a good basis could serve the description of the activities together with the video sequences of routine activities of a man provided by Bergmann et. al. [5]. Then the results of these measurements could be used in verification of mathematical models.

Mathematical model should improved as well. Different radius of the ball of the prothesis and of the acetabulum should be taken into account.

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Acknowledgement

The research is supported by the Ministry of Education project: Transdisciplinary research in Biomedical Engineering, No. MSM 210000012.