Hip stress distribution may be a risk factor for avascular necrosis of femoral head

D. Dolinar[1], M. Ivanovski[2], I. List[2], M. Daniel[3], B. Mavcic[1], M. Tomsic[4], A. Iglic[5] and V. Kralj-Iglic[2]

[1] Department of Orthopaedic Surgery, University Medical Center Ljubljana, Ljubljana, Slovenia
[2] Institute of Biophysics, Faculty of Medicine, University of Ljubljana, Ljubljana, Slovenia
[3] Technical University Kosice, Kosice, Slovakia
[4] Department of Rheumatology, University Medical Center Ljubljana, Ljubljana, Slovenia
[5] Faculty of Electrical Engineering, University of Ljubljana, Ljubljana, Slovenia

Abstract— **Avascular necrosis of femoral head (AN) is a hip disorder with various risk factors, however, the underlying mechanisms are not yet understood. In order to elucidate the effect of the mechanical factors on AN we have compared a group of hips at risk for AN and a group of healthy hips with respect to biomechanical parameters: functional angle of the weight bearing area (ϑ_F), position of the stress pole (Θ), index of the gradient of the contact stress at the lateral border of the load bearing area (G_p) and peak contact hip stress (p_{max}). The test group representing hips at risk for AN consisted of 32 male hips contralateral to the necrotic hips while the control group consisted of 46 healthy male hips. The biomechanical parameters we computed with the HIPSTRESS method (based on measurements of geometrical parameters from standard anterior-posterior pelvic radiographs). The average values of parameters pertaining to both groups were compared by the unpaired two-sided Student t-test. The functional angle of the weight bearing area was on the average larger (more favorable) in the control group (112.9º±13.5º) than in the test group (105.0º±12.4º), the difference (7%) being statistically significant ($p < 0.01$). The position of the stress pole was more lateral (less favorable) in the test group (15.44º±7.23º) than in the control group (11.80º±7.58º), the difference (27%) being statistically significant ($p = 0.037$). The index of the hip stress gradient was higher (less favorable) in the test group (-17.23º±17.16º x $10^3 m^{-3}$) than in the control group (26.05±16.85 x $10^3 m^{-3}$), the difference (40%) being statistically significant ($p = 0.028$) while we found no statistically significant difference in the peak contact stress between the two groups. Our results indicate that a less favorable steep stress distribution over a smaller load-bearing area is a risk factor in AN.**

Keywords— **Avascular necrosis, Biomechanics, Hip stress, Femoral head.**

I. INTRODUCTION

Avascular necrosis of femoral head (AN) is characterized by deterioration of the bone tissue. It represents together with osteoarthritis secondary to it a serious orthopaedic problem affecting mostly young and middle aged populations [1]. In spite of numerous studies, mechanisms leading to ischemic and necrotic processes are not yet understood. In about one third of patients the risk factors cannot be determined [2] while disorders and risk factors connected to the onset of AN include alcoholism [1], corticosteroid therapy in patients with connective tissue diseases and transplants [1], sickle cell anemia [3], HIV [2, 3], antiphospholipid syndrome [5], pregnancy [2, 6] and some others [7, 8, 9]. It was suggested that recidivant microfractures in the region of highly loaded femoral head may lead to microvascular trauma and thereby induce development of AN [10]. A question can therefore be posed whether biomechanical parameters such as stresses in the hip are important in the onset of AN.

The method HIPSTRESS enables determination of resultant hip force in the one legged stance [11] and the corresponding distribution of contact hip stress [12] in a large number of patients by using the data obtained from standard anteroposterior radiographs. The method has been validated in clinical studies [13-18]. It was shown that small functional angle of the load-bearing area ϑ_F, unfavorable distribution of the hip stress described by lateral position of the stress pole Θ, high index of gradient of contact hip stress (large slope of the distribution at the lateral border of the load bearing area) G_p and elevated peak contact stress in the hip joint p_{max} may be related to increased risk for development of hip osteoarthritis. While peak contact stress pmax is a well known and frequently used quantity, the index of stress gradient G_p [13] and the functional angle of the load-bearing area ϑ_F [14] have only recently been given attention; therefore they are briefly described below.

Index of gradient of contact hip stress G_p characterizes the slope of the distribution at the lateral border of the load bearing area. The functional angle of the load bearing area ϑ_F describes the amount of the articular sphere that is occupied by the load bearing area. The lower (more negative) index of gradient and the larger the functional angle of the load bearing area, the more favorable is stress distribution. In a population study it was shown [12] that the change of sign of G_p correlates well with the clinical evaluation of hip dysplasia, i.e. positive values of G_p correspond to dysplastic hips. The functional angle of the load bearing area ϑ_F,

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which does not critically depend on the size of the pelvis and femur was proved the most relevant in samples where there is large scattering in the size of the geometrical parameters, as for example in a group of children [14] or if there is a possibility that radiographic magnification varies considerably. In these cases the effect of p_{max} and G_p (which strongly depend on radiographic magnification) can not be envisaged due to large scattering and poor statistical significance.

It is the aim of this work to investigate the role of the above biomechanical parameters in the onset of AN.

II. Materials and Methods

From the archive of the Department of Orthopaedic Surgery, University Medical Center Ljubljana, Slovenia we selected standard anterior-posterior radiographs of pelves and proximal femora of 32 adult male persons (32 hips) who were treated due to AN between 1972 and 1991. In patients who were operated due to AN, only radiographs that were taken before the operation were used. It was assumed that prior to necrosis both hips had the same geometry. As the necrotic process had already caused changes in the geometry of some hips, the hips contralateral to the necrotic ones were considered in the study. These hips are referred to as hips at higher risk for AN. For comparison, we selected radiographs of 23 male persons (46 healthy hips) pertaining to patients who had had a radiograph of the pelvic region taken at the same institution for reasons other than hip joint disease (e.g. lumbalgia).

In our study we considered only male hips. As the values of peak hip stress importantly depend on the gender [19] it is important to have gender matched groups in statistical analysis. In our archives we did not find an adequate number of radiographs of female hips with AN that would fulfill the inclusion criteria to perform statistical analysis.

Three-dimensional biomechanical models [11, 12] were used to estimate the peak contact stress distribution given by its peak p_{max} [12], location of its pole Θ [12], index of the contact stress gradient G_p [13] and functional angle of the load-bearing area ϑ_F [14]. The input parameters of the model for the resultant hip joint force are geometrical parameters of the hip and pelvis: interhip distance l, pelvic height H, pelvic width laterally from the femoral head center C and coordinates of the insertion point of abductors on the greater trochanter (point coordinates T_x, T_z) in the frontal plane. The model of the resultant hip force is based on the equilibria of forces and torques acting between the body segments. The three-dimensional reference coordinates of the muscle attachment points were taken from the work of Dostal and Andrews [19] and scaled with regard to the pelvic parameters (l, C, H, T_x, T_z) assessed from anterior-posterior radiographs for each individual subject. In some radiographs of the patients with AN the upper part of the pelvis was not visible. In these patients the contour was extrapolated on the basis of the visible parts. As in some hips with AN the femoral head was considerably flattened superiorly, centers of rotation on both sides, corresponding to the pre-necrotic situation were determined by circles fitting the outlines of the acetabular shells.

The contact stress distribution was calculated by assuming that the hip joint consists of spherical head and hemispherical acetabular shell, the two being separated by elastic cartilage layer. When unloaded, the head and the shell are concentric while loading causes a deformation of the cartilage. The head and the shell reach closest approach in a particular point on an imaginary articular surface. This point is called the stress pole [20]. Integration of the contact stress over the load-bearing area yields the resultant hip joint force. The lateral border of the load-bearing area is defined by the coverage of the head by the acetabulum while the medial border is defined by the condition of vanishing stress [12]. The value of stress at the pole and the position of the pole on the articular sphere are calculated by solving a system of algebraic equations, one of them being nonlinear. The input parameters of the model for stress distribution are the magnitude and direction of the resultant hip joint force R and the geometrical parameters of the hip: radius of the femoral head r and Wiberg centre-edge angle ϑ_{CE}.

To describe stress distribution, we determined biomechanical parameters ϑ_F, Θ, G_p and p_{max} for each hip. The parameters p_{max} and G_p were normalized to the body weight (W_B) to extract the influence of hip geometry on stress. The average values corresponding to the test and the control group were compared by the unpaired two-sided Student t-test.

III. Results

Table 1 shows the computed biomechanical parameters: the functional angle of the lead bearing area ϑ_F, position of

Table 1 Biomechanical parameters (mean ± standard deviation) in the test group (32 hips contralateral to the necrotic hips) and in the control group (46 normal hips). Statistical significance was determined by the unpaired two-sided Student t-test. The more favorable value is marked with *.

Parameters	Test group	Control group	Difference	p value
ϑ_F [deg]	105.0 ± 12.4	112.9 ± 13.5*	7 %	<0.01
Θ [deg]	15.44 ± 7.23	11.8 ± 7.58*	27 %	0.037
G_p/W_B [$10^3 m^{-3}$]	-17.32 ± 17.16	-26.05 ± 16.85*	40 %	0.028
p_{max}/W_B [m^{-2}]	2173 ± 785	2090 ± 502*	4 %	0.604

Table 2 Geometrical parameters (mean ± standard deviation) in the test group (32 hips contralateral to the necrotic hips) and in the control group (46 normal hips). Statistical significance was determined by the unpaired two-sided Student t-test. The more favorable value is marked with *.

Parameters	Test group	Control group	Difference	p value
C [mm]	60.0 ± 10.0*	58.5 ±8.6	3%	0.463
H [mm]	163.0 ± 19.6*	162.4 ± 9.8	0.4 %	0.867
l [mm]	203.1 ± 17.5	199.6 ± 8.9*	2 %	0.305
x [mm]	12.5 ± 7.6*	7.6 ± 6.4	65 %	<0.01
z [mm]	74.7 ± 11.3*	69.7 ± 7.7	7%	0.033
r [mm]	28.5 ± 3.1*	27.7 ± 1.7	3 %	0.187
ϑ_{CE} [deg]	2173 ± 785	2090 ±502*	13 %	<0.01

the stress pole Θ, normalized index of the contact stress gradient (G_p/W_B) and normalized peak stress (p_{max}/W_B) in the test group and in the control group. Hips in the test group are on the average less favorable with respect to all three parameters ϑ_F, Θ, G_p/W_B and p_{max}/W_B. The differences in ϑ_F (7%), Θ and G_p/W_B are statistically significant ($p < 0.01$, $p = 0.037$ and $p = 0.028$, respectively) while the difference in p_{max}/W_B (4%) is not ($p = 0.604$). The results therefore show less favorable stress distribution in hips at risk for AN while there is no significant difference in the absolute values of the contact hip stress.

In order to better understand the differences in biomechanical parameters the differences in geometrical parameters were studied. Table 2 shows geometrical parameters used in the models for the above biomechanical parameters in the test group and in the control group. The center-edge angle ϑ_{CE} is smaller (less favorable) in the test group than in the control group, the difference (13%) is statistically significant ($p < 0.01$). However, the position of the insertion point of the effective muscle on the greater trochanter (both, the lateral component z and the inferior component x) is more extensive (more favorable) in the test group than in the control group. Both differences (7% and 65%, respectively) are statistically significant ($p = 0.033$ and < 0.01, respectively). Also other geometrical parameters were more favorable in the test group, however, the differences were small and statistically insignificant.

IV. Discussion and Conclusions

The shape of the stress distribution (described by ϑ_F, Θ and G_p/W_B) is on average considerably and statistically significantly different in both groups. In the test group the distribution is steeper, the pole lies more laterally, the gradient index is larger (less negative) and the functional angle of the weight bearing area is smaller than in the control group. This renders hips with increased risk for AN less favorable regarding the stress distribution, however we did not find statistically significant difference in p_{max}/W_B. Magnification of radiographs was not known as no unit with known length was visible in the picture. Magnification may vary considerably contributing to the scattering in the measured distances and this may be one of the reasons for poor statistical significance in p_{max}/W_B.

The differences in the biomechanical parameters may be explained by the difference in the geometrical parameters. The difference in pelvic height H and width C and in the interhip distance l were very small (below 3%) and statistically insignificant while the difference in the vertical coordinate of the insertion of the effective muscle on the greater trochanter (x) was statistically significant, but this parameter does not influence much the biomechanical parameters [21]. The differences in the remaining three parameters (lateral coordinate of the insertion of the effective muscle on the greater trochanter, radius of the femoral head and center-edge angle) can however contribute to the explanation of the differences in biomechanical parameters. The center-edge angle ϑ_{CE} is the most important parameter in determination of contact stress distribution. Larger ϑ_{CE} corresponds to lower p_{max}/W_B and smaller G_p/W_B. Table 2 shows that ϑ_{CE} is statistically significantly lower in the test group ($p < 0.01$) indicating that p_{max}/W_B and G_p/W_B would be higher in hips at risk for AN. However, p_{max}/W_B and G_p/W_B strongly depend also on the radius of the femoral head (p_{max}/W_B is inversely proportional to the square of r and G_p/W_B is inversely proportional to the third power of r). Although the difference in the radii of the two groups is not statistically significant ($p = 0.187$), the difference (3%) is in favor of hips in the test group. Further, the lateral position of the insertion of the effective muscle is for 7% statistically significantly larger (more favorable) in the test group than in the control group ($p = 0.033$). The effect of the smaller center-edge angle is therefore counterbalanced by the effect of larger femoral head and more laterally extended greater trochanter.

It has been hypothesized that transient osteoporosis of the bone marrow oedema syndrome may be the initial phase of osteonecrosis of the femoral head [20, 21] and that there may be a common patophysiology. Transient osteoporosis is connected to recidivant microfractures and microvascular trauma at highly loaded regions of the bone leading to the ischemia of the affected part of the bone [21]. Higher contact hip stress may increase the probability and the extent of microfractures of the affected bone thereby making the repair more difficult. Furthermore, the replicative capacity of osteoblast cells of the intertrochanteric area of the femur in osteonecrosis patients was found to be significantly reduced comparing to patients with osteoarthritis [22]. Thereby, elevated contact hip stress could accelerate the processes leading to AN.

Our study did not answer the question whether AN occurs due to common underlying mechanisms. There seems to be a heterogeneous etiology connected to different underlying mechanisms. Elevated contact hip stress with unfavorable (steep) stress distribution is yet another possible relevant mechanism which should also be considered in future investigations on onset and development of AN. Our results indicate that hips with less favorable (steeper) stress distribution are at greater risk for development of AN than hips with more uniform stress distribution.

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CORRESPONDENCE

Author:	Prof. Dr. Veronika Kralj-Iglic
Institute:	Institute of Biophysics, Faculty of Medicine, Uni-Lj
Street:	Lipiceva 2
City:	SI-1000 Ljubljana
Country:	Slovenia
Email:	vera.kralj-iglic@biofiz.mf.uni-lj.si