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The shape of acetabular cartilage optimizes hip contact stress distribution

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Abstract

The biomechanical role of the horseshoe geometry of the acetabular cartilage is described using a three-dimensional mathematical model. It is shown that the acetabular fossa contributes to a more uniform articular contact stress distribution and a consequent decrease in the peak contact stress. Based on the results it is suggested that the characteristic horseshoe shape of the articular cartilage in the human acetabulum optimizes the contact stress distribution in the hip joint.

Key words biomechanics; contact stress; facies lunata; hip joint; mathematical modelling.

Introduction

The human hip joint is a spherical joint where the spherical bone surfaces of femur and acetabulum, which are covered by cartilage, are in close contact. The cartilage covers almost all the femoral head whereas the central and inferior part of the acetabulum, i.e. the acetabular fossa, is not covered by cartilage (Petersilge, 2000). The acetabular articular cartilage therefore attains a characteristic horseshoe-shaped structure called the facies lunata.

It is generally accepted that mechanical loading is an important factor influencing development of the cartilage (see Daniel et al. 2003, and references therein). The shape of the cartilage thus influences the mechanical conditions in the hip. In previous biomechanical studies the specific shape of the articular cartilage was either not studied (Legal, 1987; Iglič et al. 2002) or fixed (Genda et al. 2001). This study was intended to estimate how the contact stress distribution in the human hip joint is influenced by the characteristic horseshoe shape of the acetabular cartilage.

Methods

To estimate the contact stress distribution in the hip joint, the resultant force transmitted between the femoral head and acetabulum ($\mathbf{R} = (R_x, R_y, R_z)$) should be known. To assess the value of **R**, a biomechanical model of the human hip was used in order to solve equilibrium equations for moments and forces acting on the pelvis in a one-legged stance (Iglič et al. 2002). The one-legged stance was chosen as a representative body position, frequently attained in everyday activities (Daniel et al. 2001).

The origin of the coordinate system was chosen in the centre of the acetabular shell so that *x* and *z* axes lie in the frontal plane and the *y* and *z* axes in the sagittal plane of the body (Fig. 1A). In the one-legged stance, the hip joint resultant force **R** lies almost in the frontal plane of the body (Iglič et al. 2002). Therefore, the force **R** can be expressed as:

$$\mathbf{R} = (R \sin \vartheta_R, 0, R \cos \vartheta_R) \quad (1)$$

where ϑ_R is the inclination of **R** with respect to the saggital plane (Fig. 1A). Using the above-mentioned

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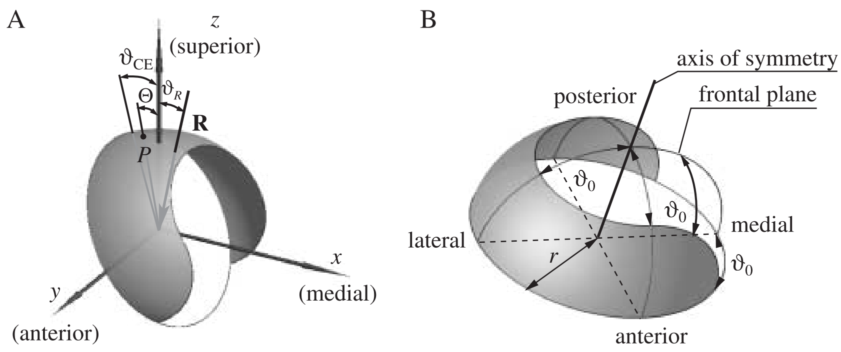

Fig. 1 Schematic presentation of the coordinate system (A) and geometrical model of the articular cartilage (B). In the model the x–z plane is identical with the frontal plane and the y–z plane with the sagittal plane. Inclination of the acetabulum (ϑ_{CE}), the direction of the force **R** (ϑ_R) and position of the stress pole P, given by angle Θ, are denoted. The angle ϑ_0 defines the size of the acetabular fossa, and r denotes the radius of the articular surface. The axis of symmetry of the hemispherical acetabular shell and the frontal plane are also denoted.

biomechanical model, it was found that for a human with reference geometry of the hip and a body weight of 800 N: $R = 2160$ N and $\vartheta_R = 5°$ (Iglič et al. 2002).

In order to calculate the contact stress distribution for given R and ϑ_R, we adapted a previously developed three-dimensional mathematical model (Daniel et al. 2001; Iglič et al. 2002). Within the model it is assumed that the spherical femoral head and hemispherical acetabulum are covered by a cartilage layer of constant thickness. When unloaded, the articular surfaces of the femoral head and acetabulum have the same radius r and are congruent. Upon loading, the femoral head is moved toward the acetabulum and the cartilage in the hip joint is squeezed. Assuming that the hip joint is well lubricated, we take into account that the tangential stresses in the articular surface of the hip joint are negligible (Legal, 1987). Hence, only compressive, i.e. radial stresses (p), are considered in our model. If the cartilage is considered as a linear elastic material, then according to Hooke's law the contact stress in the cartilage layer is proportional to displacement of the femoral head with respect to the acetabulum. Assuming a spherical shape of both joint surfaces, there is only one point where the surfaces of the acetabulum and the femur are closest. We define this point as the pole of stress distribution (P) (see Fig. 1). The deformation of the cartilage and the corresponding radial stress (p) at any point of the articular surface is proportional to the cosine of the angle between this point and the pole of the stress distribution (Brinckmann et al. 1981):

$$p = p_0 \cos \gamma \tag{2}$$

where p_0 is the stress at the stress pole. The contact stress integrated over the articular surface $\int_A p d\mathbf{A}$ should be equal to the force transmitted through the hip joint **R**:

$$\left(\int_A p d\mathbf{A} \right)_x = R_x, \quad \left(\int_A p d\mathbf{A} \right)_y = R_y, \quad \left(\int_A p d\mathbf{A} \right)_z = R_z \tag{3}$$

where **A** is the area of non-zero contact stress, i.e. the weight-bearing area. Equations (3) represent a system of three integral equations for three unknown quantities: the azimuthal angle (Φ) and polar angle (Θ), which determine the position of the stress pole, and the value of p_0. The values R and ϑ_R (see Eq. 1) are input data calculated as described above. Because the force **R** lies in the frontal plane, which is also the plane of symmetry of acetabular shell, it follows from the second of Eqs (3) that $\Phi = 0$, which means that the stress pole must lie in the frontal plane (Iglič et al. 2002). The position of the stress pole in the frontal plane, given by its inclination from the sagittal plane (Θ), and p_0 are then determined from the first and third of Eqs. (3). The angle Θ is denoted positive if the pole is on the lateral side of the frontal plane and negative if the pole is on the medial side. The weight-bearing area is bounded at the lateral side by the acetabular rim. The medial boundary of the weight-bearing area is not known in advance but is calculated from the condition of vanishing stress ($\cos \gamma = 0$). The medial border depends on the value of Θ because it consists of all points lying at an angular distance of 90° from the stress pole. The hemispherical acetabular shell is taken to be inclined at an angle ϑ_{CE} in the lateral direction (Fig. 1A), and no acetabular anteversion was taken into account.

To understand better the basic mechanism of the stress distribution in the mathematical model described above, Fig. 2 shows schematically the stress distribution in the hip joint articular surface for two

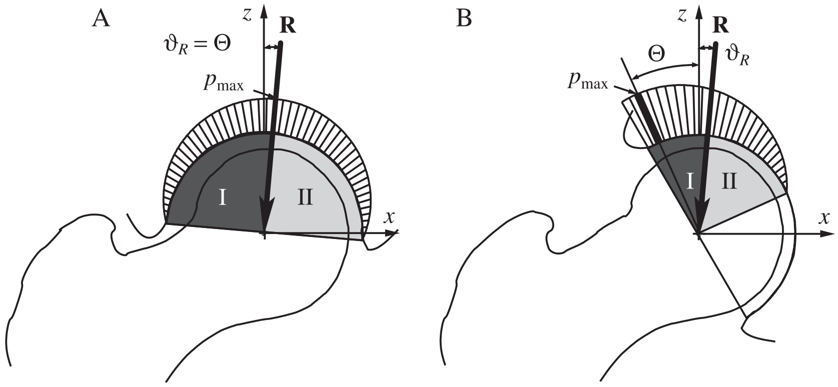

Fig. 2 Schematic figure of the stress distribution acting on the articular surface for two different rotations of the acetabulum. If the hemispherical contact articular surface is symmetric with respect to the force **R** (A), the stress pole coincides with the direction of the resultant force **R**. If the acetabulum is rotated in the medial direction (B), the stress distribution is asymmetric with respect to the direction of the resultant force (**R**) and the stress pole is moved to region I.

different orientations of the acetabulum, in both cases without the acetabular fossa. The hip joint resultant force **R** divides the acetabular surface into two parts. The lateral part is bounded by the lateral acetabular rim and by the plane of the force **R** inclined by the angle ϑ_R with respect to the sagittal plane of the body (area I in Fig. 2), whereas the medial part consists of the rest of the acetabular surface (area II in Fig. 2). The sum of the resultant force of the contact stress transmitted through the lateral part of the acetabular surface and the resultant force of the contact stress transmitted through the medial part of the acetabular surface should be equal to the total resultant force **R**. If the available areas of both parts of the acetabular surface are equal, the direction of the stress pole coincides with the direction of the force **R** (Fig. 2A). However, in the normal hip, the acetabulum is rotated in the medial direction (Fig. 2B) with respect to the situation presented in Fig. 2(A) and therefore the available weight-bearing area of the lateral part (area I) is smaller than the weight-bearing area of the medial part (area II) (Fig. 2B). Because the integral of the contact stress over the complete weight-bearing area must give the resultant hip force **R** (Eqs 3), the contact stress should on average be higher in region I than in region II of the weight-bearing surface (Fig. 2B). In the case of the cosine stress distribution function (Eq. 2), this means that the pole of the stress distribution should be located laterally with respect to the direction of force **R**.

The fact that the direction of the peak contact stress does not coincide with the direction of the hip joint resultant force **R** has also been confirmed in different experimental studies (Brown & Shaw, 1983; Hodge et al. 1989). As shown in Fig. 2B, the lateral position of the stress pole with respect to the direction of **R** provides a non-uniform stress distribution. A strongly non-uniform stress distribution is evident in dysplastic hips where the small lateral coverage of the femoral head by the acetabulum strongly increases the above-mentioned asymmetry in areas I and II (Pauwels, 1976; Mavčič et al. 2002).

In the model used hitherto (Iglič et al. 2002), it was assumed that the whole hemispherical shell of the acetabulum is covered by cartilage. Within the present work the geometry of the acetabular cartilage was modified by introducing a non-weight bearing area on the hemispherical shell of the acetabulum. This non-weight-bearing area represents the cartilage-free region of the acetabulum, i.e. the acetabular fossa (Fig. 1B).

In our mathematical model, the acetabular fossa is taken to be symmetrical with respect to the frontal plane. The frontal plane of the body is also a plane of symmetry of the acetabular shell, dividing the acetabulum into anterior and posterior parts (Fig. 1B). Another plane perpendicular to the frontal plane and parallel to the sagittal plane divides the acetabular shell into a lateral part and a medial part (Fig. 1B). In the lateral part the circular rim of the acetabular fossa consists of all points that have angular distance ϑ_0 from the axis of symmetry of the complete hemispherical acetabular shell (Fig. 1B). In the medial part of the acetabular shell, the rim of the acetabular fossa consists of all points that have a constant angular distance ϑ_0 from the frontal plane (Fig. 1B). The above parametric definition of the rim of the acetabular fossa allows us to model the typical horseshoe shape of cartilage in the hip joint articular surface. For a chosen value of ϑ_0 the solution of the above integral Eqs. (3) then gives us the values of contact stress at the pole (p_0) and the position of the stress pole, determined by the angle Θ. If the stress pole lies inside the weight-bearing area then the peak contact stress (p_{max}) is equal to p_0. If this is not the case, the point of maximal stress is the point on the rim of the weight-bearing area that is closest to the stress pole.

Results

Taking into account the above model for the geometry of the cartilage layer in the hip joint, we studied the dependence of the hip joint peak contact stress (p_{max}), the position of the stress pole (Θ) and the size of the weight-bearing area (A) on the size of the acetabular fossa determined by the angle ϑ_0 (Fig. 3). It can be seen in Fig. 3 that p_{max} decreases with increasing ϑ_0 until it reaches its minimum at $\vartheta_0 = 29°$. Further increase in the size of the acetabular fossa is no more favourable because it considerably increases the peak contact stress (Fig. 3A). Figure 3 also shows that the pole of the stress moves medially while the size of the weight-bearing area decreases with increasing values of ϑ_0 (Fig. 3B,C).

In Fig. 4, the contact stress distributions in the frontal plane and the size of the weight-bearing area for three selected sizes of the acetabular fossa are shown schematically. The projection of the weight-bearing area on the frontal plane is marked by cross-hatching. The case when $\vartheta_0 = 0°$ corresponds to acetabular cartilage that has a hemispherical shape, i.e. the cartilage layer without the acetabular fossa (Fig. 4A). It can be seen in Fig. 3A that without the acetabular fossa the contact stress distribution would be less uniform with maximal stress located close to the lateral acetabular margin. The predicted uniform stress distribution due to the acetabular fossa is in good agreement with the contact stress distribution estimated from the density of the bone structures in radiographs of healthy hips (Pauwels, 1976).

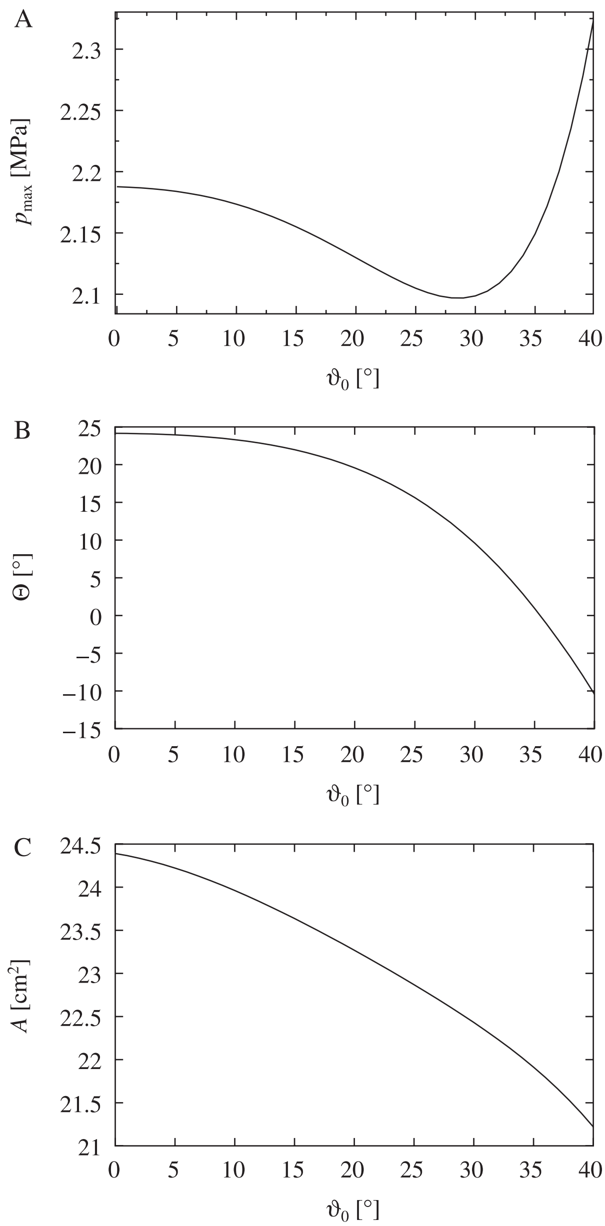

Fig. 3 Dependence of the peak contact stress p_{max} (A), the position of the stress pole Θ (B) and the size of the weight-bearing area A (C) on the size of the acetabular fossa determined by the angle ϑ_0. The values of the model parameters are: R = 2000 N, $\vartheta_R = 5°$, $\vartheta_{CE} = 30°$ and r = 2.7 cm.

Discussion and conclusions

The contact stress distribution in the human hip is considered to be an important factor influencing the development of the hip. In the present work, we have shown that the specific horseshoe shape of articular cartilage in the human hip joint decreases the peak contact stress (Fig. 3A) and renders the stress distribution more uniform (Fig. 4).

The influence of the acetabular fossa on the contact stress distribution can be viewed as the interplay of the influence of two parameters: the direction of the resultant force **R** and the geometry of the articular cartilage. The acetabular fossa represents the part of the articular surface that cannot bear weight. It is located medially with respect to the direction of the resultant force **R**, i.e. in region II (Fig. 2B). If the area of the acetabular fossa is not too large, the acetabular fossa renders the weight-bearing area more symmetric with respect to the direction of the resultant force **R**. The

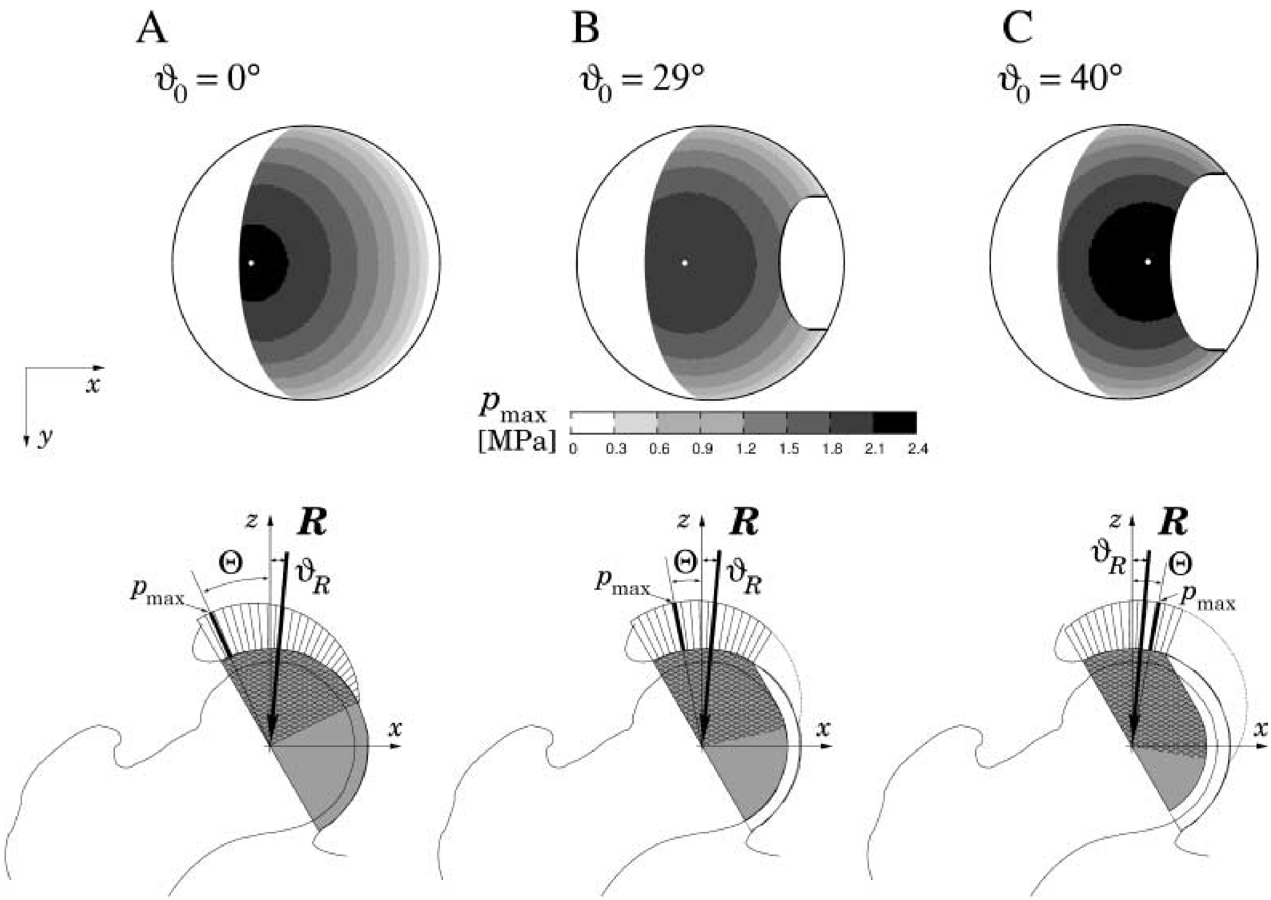

Fig. 4 The contact stress distribution projected on the transverse plane (upper figures) and on the frontal plane (lower figures) for various sizes of the acetabular fossa determined by the angle ϑ_0. In the upper figures, the position of the stress pole is denoted by the white spot. In the lower figures, the grey area is the projection of the cartilage on the frontal plane, and the white area in the acetabulum represents the acetabular fossa. The weight-bearing area is marked by cross-hatching. The position of the peak contact stress (p_{max}) and stress pole (Θ) are denoted. The values of the model parameters are as in Fig. 3.

increased symmetry of the articular surface due to the acetabular fossa moves the stress pole medially towards the force **R** (Fig. 3B). Consequently, the weight-bearing area is also extended in the medial direction (Fig. 4). However, such spreading of the weight-bearing area medially is not large enough to compensate for the reduction in the weight-bearing area due to the non-weight-bearing area of acetabular fossa (Fig. 4). Therefore, the effect of stress decrease due to the acetabular fossa cannot be explained as an outcome of the possible increase in the weight-bearing area (see also Fig. 3C). To explain the decrease of stress due to the acetabular fossa, the stress redistribution from the acetabular fossa should be considered (Fig. 4).

Figure 4B shows that the predicted medial shift of the stress pole (white spot) after an increase in the size of the acetabular fossa increases the region of high stress around the stress pole, as the stress pole was moved from the lateral acetabular rim. Hence, the contact stress is distributed more uniformly (Fig. 4B) and the peak contact stress is decreased (Fig. 3A). However, further increase in the size of the acetabular fossa (i.e. of the angle ϑ_0) moves the pole of stress closer to the medial acetabular rim (i.e. closer to the rim of the acetabular fossa). Hence, the available weight-bearing area around the stress pole is reduced in size again (Fig. 4C) and the peak contact stress is consequently increased (Fig. 3A).

To conclude, the horseshoe geometry of the articular cartilage contributes to a lower and more homogeneously distributed hip joint contact stress, which may be important in various ways. It has been suggested that long-term elevated contact stress may cause damage to the cartilage (Hadley et al. 1990; Daniel et al. 2003) and therefore the decrease in stress due to the acetabular fossa may contribute to prevention of cartilage damage. It has also been suggested that variation of the magnitude of the contact stress distribution over the articular surface may be even more important for cartilage longevity than the value of the peak stress (Brand et al. 2001; Pompe et al. 2003). The effect of stress redistribution caused by the specific horseshoe shape of the articular cartilage can therefore play an important role in cartilage development and longevity.

The articular cartilage is an avascular structure for which the process of loading and unloading and consequent flow of the interstitial fluid is crucial in nutrition (Nordin & Frankel, 1989). The existence of the acetabular fossa increases the percentage of the loaded articular cartilage (Fig. 4), i.e. the proportion of the cartilage where nutrition is ensured.

In the mathematical model described, several simplifications were used that could influence the accuracy of the calculated contact stress distribution. For example, in our model the femoral head and acetabulum are taken to be spherical. In normal hips the femoral head and the acetabulum are actually out-of-round by 1–3 mm. It has be shown that in the case of an ellipsoidal articular surface with the semi-axes r and $r + \Delta r$, the cosine stress distribution function (Eq. 2) can be modified by taking into account the perturbation of the first order in $\Delta r/r$, which yields the stress distribution function in the form $p = p_0 \cos\gamma(1 + 3(\Delta r/r)\sin^2\gamma)$ (Ipavec et al. 1999). Our mathematical model derivation also assumes that the cartilage layer has constant thickness and mechanical properties. The contact stress was assumed to be proportional to the cartilage deformation δ. If the properties of the cartilage vary along the articular surface, the contact stress at a given point also varies, and this is not taken into account in our model. The spatial variations in the mechanical properties and thickness of the cartilage modify the local values of the stress, which may explain the considerable variation in the hip contact stress distribution measured in different experimental studies (Brown & Shaw, 1983; Afoke et al. 1987).

The mechanics of the cartilage layer in the hip joint obviously cannot be fully described as a homogeneous continuum and a linear elastic material. In order to approach a more realistic description of the stress and force distribution in the cartilage layer, one should take into account the specific molecular structure of the joint articular surface where the two glycoprotein monolayers are adsorbed on the cartilage of both contact surfaces (Nordin & Frankel, 1989). The mechanics of this structure could be realistically described only by using the methods of theoretical physics on the molecular level developed in the field of polymer physics and the statistical physics of interfaces (see, for example, Butt et al. 2003).

During different activities the hip joint resultant force can attain various directions and magnitudes (Bergmann et al. 2001), whereas in this study only a static one-legged stance was considered. The study of the influence of the acetabular fossa on the contact stress distribution should therefore be evaluated in future studies during other activities besides a one-legged stance, such as walking up a set of stairs.

The accuracy of the contact stress distribution predicted here could certainly be improved by removing some of the above-mentioned simplifications assumed in our mathematical model. However, we believe that the main conclusions of our work, i.e. that the acetabular fossa contributes to a lower and more uniform stress distribution in the hip joint, would not be changed by using a more advanced mathematical model. The specific shape of the facies lunata can therefore be considered as an important factor that optimizes the contact stress distribution in the hip joint.

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