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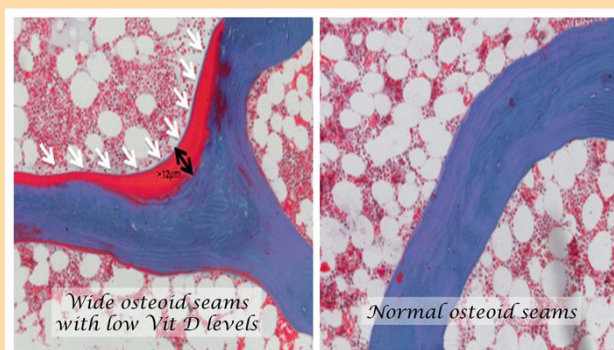
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## Polyethylene Wear is Related to Patient-specific Contact Stress in THA

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### Abstract

**Background** General numerical models of polyethylene wear and THA simulators suggest contact stresses influence wear. These models do not account for some patient-specific factors. Whether the relationship between patient-specific contact stress and wear apply in vivo is unclear.

**Questions/purposes** We therefore determined whether (1) contact stress distribution at the prosthesis-cup interface

and (2) hip geometry and cup inclination are related to wear in vivo.

**Methods** We retrospectively reviewed the radiographs of 80 patients who had aseptic loosening of their THAs as determined by radiographic criteria. We determined linear penetration and volumetric wear using postoperative and last followup radiographs. Contact stress distribution was determined by the HIPSTRESS method. The biomechanical model was scaled to fit the patient's musculoskeletal geometry of the pelvis, trochanteric position, and cup inclination using the standard postoperative radiograph.

**Results** Linear penetration and volumetric wear correlated with peak contact stress. Polyethylene wear was greater in THAs with a medial position of the greater trochanter and smaller inclination of the acetabular cup.

**Conclusions** Our observations suggest wear is specific to contact stresses in vivo.

**Clinical Relevance** Long-term wear in a THA can be estimated using contact stress analysis based on analysis of the postoperative AP radiograph.

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Each author certifies that his or her institution approved the human protocol for this investigation, that all investigations were conducted in conformity with ethical principles of research, and that informed consent for participation in the study was obtained.

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### Introduction

Osteolysis induced by polyethylene wear particles is one of the major causes of long-term failure in total joint arthroplasties [9]. The wear in a THA depends on the type of implant, the patient, and surgical factors that interact in complex ways [4].

Several studies have attempted to explain wear mechanisms using finite element [11] or elasticity analysis [33, 34]. These models apply the Archard/Lancaster relationship to calculate the wear coupling behavior assuming that wear is proportional to contact stress, sliding speed, and a bearing

couple-dependent wear factor [11]. Model simulations focus on estimating the effects of various prosthesis designs on polyethylene wear. The models suggest the geometry of the prosthesis, ie, the radius of the femoral head [25], polyethylene cup thickness [22], or cup inclination [33, 34] and the contact surface properties, ie, femoral head roughness [5], influence wear. The wear rate assessed using these mathematical models [25, 33, 34, 38] agrees well with the wear measured in a clinical study [20], suggesting an important role of contact stress in mechanical damage of the polyethylene surface. One study using a hip simulator [27] found increased total load related to greater wear, again suggesting a role of contact stress in wear. However, hip simulator tests are based on simplified loading cycles in comparison to the actual in vivo load [26].

Clinical studies have documented differing directions of penetration of the head into the cup among patients with the same type of prosthesis [3, 43], indicating the importance of patient-specific contact stress in polyethylene wear. Wear also depends on activity and hip geometry [35]. To confirm the relationship between wear and contact stress in vivo, greater wear should be observed in patients with higher contact stress at the prosthesis-cup interface. Using radiostereometry, The et al. [40] found a positive correlation between the in vivo peak contact stress and linear wear 1 year postoperatively, but no correlation at 2 and 5 years postoperatively. However, the findings were not conclusive as the number of patients in this study was relatively small (31), the followup was far less than the average lifetime of the prosthesis, and the amount of wear was relatively small. To confirm whether stress correlates with wear, we designed a retrospective study including a larger group of patients assessed for stress postoperatively and at prosthesis revision for wear.

We asked specifically (1) whether wear correlated with peak contact stress in vivo; and (2) whether wear correlated with trochanteric position, pelvic dimensions, and trochanteric cup placement.

## Patients and Methods

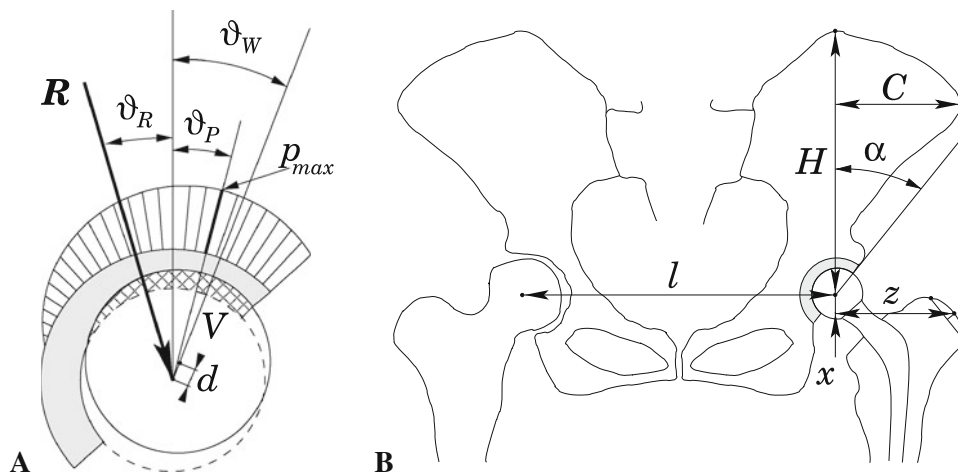
We retrospectively reviewed the radiographs of 80 patients (43 women and 37 men) who had revision hip arthroplasty for aseptic loosening between 1997 and 2001. Their median age at revision was 82 years (range, 57–92 years). The median period between the first arthroplasty and revision was 12 years (range, 4–17 years). Their mean weight at revision was 73.8 kg (SD, 10.6 kg; range, 52–92 kg). The study did not require approval by the local ethical committee but it was performed in accordance with the ethical standards of the 1964 Declaration of Helsinki as revised in 2000.

In vivo polyethylene wear depends on the size of the femoral head and the material of the contact couple [9] with minor effects of the design of the femoral component [18]. To study the general effect of femoral head and material on wear, we chose patients with implants produced by various manufacturers (Link, Hamburg, Germany; DePuy, Warsaw, IN, USA; Protek, Bern, Switzerland; Aesculap, Tuttlingen, Germany; Allo Pro, Baar, Switzerland), but with identical bearing couples regarding material and geometry. A prosthesis with a CoCrMo head and an ultrahigh molecular weight polyethylene (UHMWPE) cup with an inner surface diameter of 32 mm was implanted in all patients. All acetabular cups were cemented. The wall thickness of all cups was greater than 6 mm.

Polyethylene wear was assessed by one observer (RK) using the method of Livermore et al. [22] and a computer-assisted technique [21] that reportedly provides an accurate estimation of wear [12]. Two AP pelvis radiographs of each patient were selected for the study; the first was obtained immediately after the initial surgery, and the second was made immediately before the revision arthroplasty. We assumed the 3-D wear vector is projected onto the 2-D plane of the radiograph [25]. For cups with little anteversion, the wear is assumed to lie in the frontal plane [39], as suggested by Hui et al. in experimental measurements of retrieved liners [12]. Therefore, only cups with acetabular anteversion less than 20° were included in the study [39]. Both radiographs were corrected for magnification using the diameter of the femoral head as a reference and centered on the position of the acetabular component [21]. The displacement of the center of the prosthesis heads in the postoperative and followup radiographs was taken as the linear penetration  $d$  (Fig. 1). This technique does not assume that the femoral head center and the acetabular center are initially identical [6] and reportedly has an accuracy of 0.075 mm [26], with low intraobserver and interobserver variability (intraclass correlations [ICC], 0.97 and 0.96) [10]. The angle between the direction of displacement of the centers of the femoral heads with respect to the sagittal plane was taken as the direction of wear ( $\vartheta_w$ ) (Fig. 1A). In the plane of the radiograph, the direction of wear was considered positive in the lateral direction from vertical and negative in the medial direction (Fig. 1). The angle of the acetabular cup inclination  $\alpha$  was measured (Fig. 1B) and the volumetric wear ( $V$ ) was computed using the modified formula of Kořak et al. [21] as described by Ilchmann et al. [16]:

$$V = r^2 d / 2 (\pi + \pi \sin \beta + d \sin(2\beta) / r)$$

where  $r$  is the radius of the femoral head and  $\beta$  is equal to  $(\alpha - \vartheta_w)$ .



**Fig. 1A–B** (A) A schematic view of the acetabular cup shows penetration of the femoral head into the polyethylene cup. The direction of linear penetration ( $\vartheta_W$ ), the direction of the vector to the point of the peak contact hip stress ( $\vartheta_P$ ), and the direction of the hip

resultant force  $R$  ( $\vartheta_R$ ) are shown. Also shown is distribution of the contact hip stress. (B) The geometric parameters of the hip and pelvis needed to determine the peak contact hip stress ( $p_{max}$ ) on the weightbearing area by the HIPSTRESS method are shown.

The hip reaction force and the contact stress distribution on the inner surface of the acetabular cup during single-leg stance were determined for each patient individually according to the HIPSTRESS method [13–15] from the postoperative AP radiograph. Single-leg stance was taken as a representative body position because the contact stress distribution during this stance reportedly resembles the averaged contact stress distribution during the walking cycle [8].

Assessment of the hip reaction force  $R$  was based on force and torque equilibrium of the body in single-leg stance [13, 15]. The geometry of the musculoskeletal model of the hip was scaled for each patient individually according to the following geometric parameters measured from the AP radiograph (Fig. 1): height of pelvis  $H$ , width of pelvis  $C$ , and vertical and horizontal positions of the greater trochanter  $x$  and  $z$ , respectively. The resultant hip force  $R$  determined in single-leg stance lies near the frontal plane of the body [15] and can be expressed by its magnitude  $R$  and by its inclination in the frontal plane with respect to the vertical plane  $\vartheta$  (Fig. 1). Angle  $\vartheta_R$  is considered positive in the lateral direction from the vertical axis and negative in the medial direction from the vertical axis.

In addition to the resultant hip force  $R$ , input parameters of the mathematical model for calculation of contact stress [15, 17] were the geometry of the prosthesis given by the radius of the acetabular cup  $r$  and the position of the acetabular cup. We assumed no clearance between the femoral head and the acetabular cup. The position of the acetabular cup on the AP radiograph was determined by its inclination with respect to vertical  $\alpha$  (Fig. 1A), where no anteversion was assumed.

After loading, the modeled femoral head and acetabular cup were brought into contact. Because the stiffness of UMWPE is approximately two orders of magnitude smaller than that of the CoCrMo femoral head, the metallic head acts as a rigid indenter of the polyethylene [37]. We assumed that the back of the polyethylene insert was not deformed [42]. As a result of the spherical geometry of the contact surfaces, the radial deformation of the acetabular cup is [17]

$$\delta = \delta_0 \cos \gamma$$

where  $\delta_0$  is the displacement of the center of the femoral head and  $\gamma$  is the angle between the direction of the femoral head displacement and the direction of the radius vector from the center of the femoral head to the chosen point at the contact surface. The point where the deformation  $\delta$  equals  $\delta_0$  ( $\gamma = 0$ ) is denoted as the stress pole ( $P$ ). The position of the stress pole is given in the frontal plane by the angle  $\vartheta_P$ . Angle  $\vartheta_P$  is considered positive in the lateral direction from the vertical axis and negative in the medial direction from the vertical axis (Fig. 1A). If we assume that the contact stress  $p$  is proportional to the radial strain of the polyethylene cup [24, 34], it follows that [40],

$$p = p_0 \cos \gamma$$

where  $p_0$  is the contact stress at the stress pole. The sum of the contact stresses over the contact surface is equal to the force  $R$  [7].

$$R = \int p dA$$

Solution of the integral equation yields the position of the stress pole ( $\vartheta_P$ ) and the magnitude of stress at the pole

( $p_0$ ) [7, 17]. If the pole lies within the weightbearing area, the peak stress borne by the hip is equal to its value at the pole ( $p_{max} = p_0$ ). If the stress pole lies outside the contact area, the peak stress is equal to the value of the stress at the closest point of the contact area to the pole. The values of the peak stress  $p_{max}$  and the resultant hip force  $R$  were normalized to the body weight  $W_B$  for each subject (so they are presented as  $p_{max}/W_B$  and  $R/W_B$ , respectively) to eliminate the effect of body weight variation during the followup period.

The relationship between the measured linear and volumetric wear ( $d$  and  $V$ , respectively) and the computed peak contact stress estimated from postoperative radiographs was evaluated using linear regression analysis. Linear regression also was used to test the correlation between wear characteristics  $d$  and  $V$  and the measured pelvic dimensions  $l$ ,  $H$ ,  $C$ , trochanteric position  $z$ ,  $x$ , and acetabular cup inclination  $\alpha$  (Fig. 1B). The statistical analysis was performed with GNU Octave software, NaN statistical package (JW Eaton, Version 3.0.5; Madison, WI, USA).

**Results**

We observed correlations between linear penetration  $d$  ( $r^2 = 0.70$ ;  $p < 0.001$ ) and volumetric wear  $V$  determined at follows-up ( $r^2 = 0.50$ ;  $p < 0.001$ ), with the peak contact stress  $p_{max}/W_B$  determined on the basis of the postoperative radiographs (Table 1). The magnitude of the hip reaction force  $R$  correlated with measured linear and volumetric wear ( $r^2 = 0.35$  and  $0.51$ , respectively;  $p < 0.001$ ). The position of the peak contact stress  $\vartheta_P$  coincided with the measured direction of linear penetration  $d$  (Fig. 2). In contrast, the direction of the hip reaction force pointed more medially compared with the wear direction (Fig. 2).

**Table 1.** Pearson's correlation coefficients ( $R^2$ ) and p values of the correlation between radiographic and biomechanical parameters and linear penetration ( $d$ ) and volumetric wear ( $V$ )

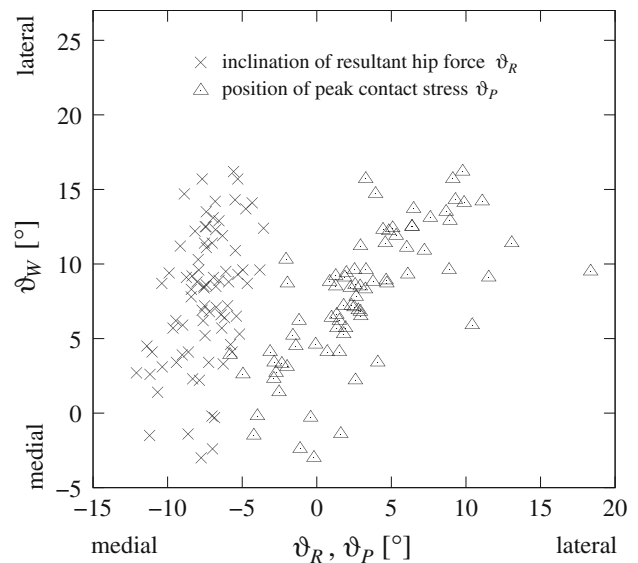
Parameter	Unit	Mean $\pm$ SD	d		V	
			$r^2$	p	$r^2$	p
$p_{max}/W_B$	kPa/N	$5.89 \pm 6.36$	0.70	< 0.001	0.50	< 0.001
$R/W_B$	N/N	$2.87 \pm 0.23$	0.35	< 0.001	0.51	< 0.001
$l$	cm	$18.42 \pm 1.05$	0.00	0.746	0.01	0.334
$H$	cm	$12.82 \pm 0.99$	0.01	0.390	0.00	0.631
$C$	cm	$5.54 \pm 0.80$	0.10	0.003	0.11	0.002
$x$	cm	$1.50 \pm 0.69$	0.00	0.877	0.01	0.398
$z$	cm	$5.33 \pm 0.67$	0.29	< 0.001	0.39	< 0.001
$\alpha$	°	$47.50 \pm 6.05$	0.43	< 0.001	0.06	0.027
$W_B$	N	$73.78 \pm 10.64$	0.03	0.123	0.01	0.284

We also observed correlations between the measured wear characteristics  $d$  and  $V$  and the mediolateral position of the greater trochanter  $z$  and acetabular cup inclination  $\alpha$  (Fig. 1B). Greater wear was related to the acetabular cup being opened more laterally, ie, small  $\alpha$ , and to the trochanter being located more medially, ie, small  $z$ . The pelvic dimensions  $l$ ,  $H$ ,  $C$ , and vertical position of the greater trochanter  $x$  (Fig. 1B) did not correlate with wear characteristics  $d$  and  $V$ .

**Discussion**

The relationship between stress and wear of a THA has been established in theoretical models and laboratory measurements; however, a 5-year followup study of THAs in a group of patients [40] yielded indecisive results. In addition to postoperative geometry, there might be other factors that influence wear such as the activity of the patient and biocompatibility of the prosthetic material. We therefore asked (1) whether wear correlated with estimated contact stress in vivo, and (2) whether wear correlated with trochanteric position, pelvic dimensions, and trochanteric cup placement.

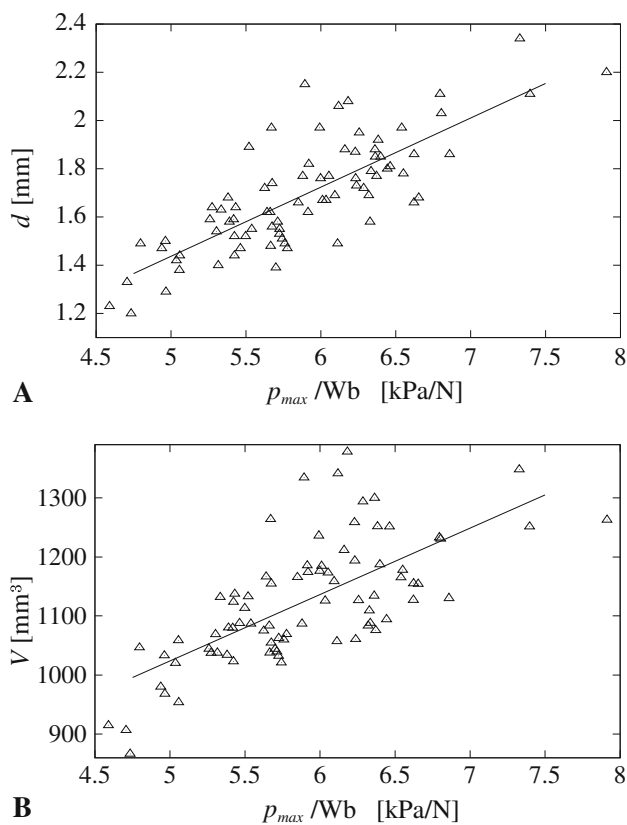
Our study has some assumptions and limitations. First, the study potentially is biased by the selection of patients who had revision hip arthroplasty for aseptic loosening. In these patients, the wear rate and corresponding contact stress can be much greater than in asymptomatic patients [9, 18]. The correlation between the contact stress and wear



**Fig. 2** The vector to the position of the peak contact hip stress  $p_{max}$  ( $\vartheta_P$ ) coincides with the direction of measured maximal linear wear  $\vartheta_W$ , whereas the direction of the hip resultant force  $\vartheta_R$  is located more medially compared with the direction of measured maximal linear wear  $\vartheta_W$ .

in asymptomatic patients can be expected to be lower owing to smaller wear rate and greater interindividual variation [18]. Second, the scatter in our data shows additional factors have a considerable influence on the magnitude (Fig. 3) and direction (Fig. 2) of THA wear. Third, the model considers single-leg stance represents all loading positions. During routine activities, the hip force attains various directions and magnitudes [32] that will change the contact stress pattern [23, 34]. Also, the effect of velocity friction during motion [1], the effect of cross-shear [20], and differences in physical activity between patients [28] are not explicitly included. To include these motion-related effects for an individual patient, additional motion analysis accompanied by dynamic hip force assessment would be required [25]. The current study was designed as a retrospective study based on analysis of radiographs. Additional dynamic hip force assessment of primary THA was not available for patients after THA revision. Fourth, we did not consider activity levels or cumulative loading in these patients. Clearly these would vary considerably and would likely account for some of the variations. Patients with high contact stress in single-leg stance would be expected to have high stresses in dynamic

loading as well. Including the motion-related wear estimation in future studies might provide better correlation between the stress and wear [1, 20]. Nevertheless, the observed correlations between parameters of contact hip stress and measured wear provide additional evidence that contact stress is an important parameter in the development of the hip and also that the HIPSTRESS method is a relevant and useful clinical method for analyzing populations and predicting wear outcomes. Fifth, in the analysis of contact stress, we assume no clearance between the femoral head diameter and the diameter of the acetabulum. Numerical simulations indicated an initial radius mismatch of the order of a few tenths of a millimeter (the current industrial tolerance range in THA [4]) has minimum consequences in the long term, because the polyethylene surface is reshaped by wear in the first hundred thousand cycles to match the radius of the head [25]. In addition, if no clearance is assumed, the proposed model [15] and the finite element models [4] predict that the peak contact stress is independent of the elastic modulus of polyethylene. This simplifies the calculation, because the exact material properties of the implanted polyethylene cup usually are unknown. Conventional UHMWPE cups were implanted in all THAs included in this study. Modern crosslinked polyethylenes exhibit considerably lower wear related to contact stress [36] and other factors like third-body wear [5] might be more important for its wear in vivo. Sixth, current contact stress modeling is based on the assumption of elastic deformation of polyethylene. The simple elasticity theory might overestimate the contact pressure resulting from plastic deformation and creep in high contact stress regions [38]. However, most contact stresses observed in our study (Fig. 3A) are below the critical value of 8 MPa [38] related to UHMWPE plastic deformation. Seventh, wear was estimated from standard AP radiographs. In comparison with the more accurate radiostereometric analysis used in a previous study [40], we were limited because measuring beads were not implanted in the patients analyzed. The use of AP radiographs is limited by the fact that a 3-D wear vector is projected onto a 2-D image, therefore the estimation is justified only for cups with small anteversion [26]. However, this was taken into account by excluding cups with anteversion greater than 20°. The method can be further improved by considering a correction procedure suggested by The et al. [39]. Eighth, the measurement of femoral head penetration  $d$  cannot differentiate between bedding-in (consisting of creep of polyethylene and settling of the liner) and true wear. Further, a more realistic estimation of wear requires multiple wear measurements at different followups [26]. Finally, cemented acetabular cups were included in the analysis. The presence of a metal shell might change the stiffness of the acetabular component, but



**Fig. 3A–B** (A) Linear penetration  $d$  and (B) volumetric wear  $V$  of the polyethylene cup determined at followup in 80 THAs correlate with the peak contact hip stress normalized to the body weight  $p_{max}/W_B$  determined on the basis of postoperative radiographs.

it is not likely to change the contact mechanics because no difference in wear between a cemented and an uncemented cup was found in a clinical study [33] and in hip simulator testing [36]. All implanted prostheses have the same femoral head diameter and polyethylene liners with a thickness of 6 mm or greater. For polyethylene liner thickness greater than 6 mm, the contact stress is less sensitive to changes in UHMWPE thickness [2] and an elastic model can be used to describe contact mechanics. The contact stress increases nonlinearly for polyethylene liner thickness less than 4 mm [2]. Therefore, the correlation of contact stress estimated by the presented method and penetration might not be straightforward in thinner polyethylene liners [2]. Wear also depends on the femoral head radius. Hips with larger contact radii exhibit higher sliding velocities increasing the wear [25], although the contact stress magnitude is lower owing to a larger contact area. As a result, hips with a larger radius have larger volumetric wear [18]. The effect of patient-specific contact stress in THAs with various geometries should be studied further.

Our data suggest polyethylene wear correlates with peak stress at the prosthesis head-cup interface. The role of contact stress in the wear process of a THA is manifested by correlations between the values of the peak contact stress  $p_{max}$  determined from postoperative radiographs and linear penetration  $d$  and volumetric wear  $V$  measured at the last followup before revision of the prosthesis. These results complement previous laboratory measurements, which also suggest a correlation between contact stress and polyethylene wear using idealized loading in joint simulators [27]. In addition to quantitative agreement between the contact stress and wear, we observed qualitative agreement between the wear direction and peak stress direction. Several studies have noted a mismatch between superolateral wear of the acetabular cup and the medial direction of the hip reaction force [19, 29, 30, 41]. The mathematical model used in this study explains why the direction of penetration coincides with the direction of the peak stress pole and not with the direction of the hip reaction force. Namely, distribution of the resultant hip force  $R$  over the weightbearing area should be taken into consideration. As the acetabulum is opened laterally, the hip loading force acts closer to the superior acetabular rim than to the inferior. This asymmetric loading force generates nonuniform stress distribution with the stress pole located laterally [7]. The closer the force acts to the superior acetabular rim, the more laterally the stress pole is positioned to satisfy equilibrium between hip force and contact stress. The material of the cup is loaded mainly in the region of high stress and this determines the superolateral wear direction.

We found the medial position of the greater trochanter is related to increased wear. From a mechanical point of view, it is known that medialization of the greater trochanter decreases the effective moment arm of the abductor

muscles, increases the muscle force, and therefore increases the hip force [31]. According to the Archard/Lancaster relationship, a higher loading force is related to higher wear [1], as we also observed in this study (Table 1). Our observations confirm those of several clinical and theoretical studies suggesting increased wear is related to the lateral inclination of the acetabular cup [30, 33, 34, 41]. A small lateral coverage of the femoral head by the acetabulum decreases the contact area and contributes to a nonuniform stress distribution with elevated peak contact stress [14, 15, 34]. This effect is similar to the increased peak contact stress in dysplastic hips resulting from reduced lateral coverage of the acetabulum [23].

The relationship between postoperative contact stress and wear at followup suggests contact stress might be used as an objective function in computer-assisted preoperative planning. Currently, only the geometry of the proximal femur and acetabulum are considered in preoperative planning (eg, TraumaCad; Columbia, MD, USA) and contact stress distribution or loading force is not considered. The peak contact stress calculation we describe is based on analysis of the whole pelvis. Even if some geometric parameters (eg,  $l$ ,  $H$ ,  $C$ ,  $x$  [Table 1]) do not show a correlation with wear, they might be important in contact stress calculation. However, a primary role for contact stress is suggested by the considerably higher value of the correlation coefficient in comparison to the correlation coefficients of the individual geometric parameters ( $z$ ,  $\alpha$ ). Prostheses with a ceramic femoral head and cross-linked polyethylene might be used in patients with unfavorable geometry of the hip to reduce the expected high wear, whereas positions of the acetabular cup related to high contact stress can be identified before implantation.

Our data confirm the role of in vivo contact stress in polyethylene wear in THA suggested in previous theoretical [2, 20, 24, 25, 33, 34] and experimental [27] studies. The relationship between postoperative contact stress and wear at followup indicates that the approach might be used for long-term wear predictions based on analysis of the standard postoperative AP radiograph.

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