



Patient-specific hip geometry has greater effect on THA wear than femoral head size



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ARTICLE INFO

Article history:

Accepted 21 October 2016

Keywords:

UHMWPE

Linear penetration

Contact pressure

Design

ABSTRACT

In vivo linear penetration in total hip arthroplasty (THA) exhibits similar values for 28 mm and 32 mm femoral head diameter with considerable variations between and within the studies. It indicates factors other than femoral head diameter influence polyethylene wear. This study is intended to test the effect of patient's individual geometry of musculoskeletal system, acetabular cup orientation, and radius of femoral head on wear. Variation in patient's musculoskeletal geometry and acetabular cup placement is evaluated in two groups of patients implanted with 28 mm and 32 mm THA heads. Linear wear rate estimated by mathematical model is 0.165–0.185 mm/year and 0.157–0.205 mm/year for 28 and 32 mm THA heads, respectively. Simulations show little influence femoral head size has on the estimated annual wear rate. Predicted annual linear wear depends mostly on the abduction angle of the acetabular cup and individual geometry of the musculoskeletal system of the hip, with the latter having the greatest effect on variation in linear wear rate.

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1. Introduction

Wear of polyethylene is considered to be a limiting factor in determining the longevity of total hip arthroplasty (THA) (Dowd et al., 2000; Takenaga et al., 2013). The amount of wear particles can be assessed from plain radiographs by measuring head penetration into acetabular cup (Ilchmann et al., 2008). Risk parameters that predispose to wear include implant design, bearing surface material, operative and patient-specific parameters (Brown et al., 2008). The diameter of the femoral head is considered to be an important design parameter related to wear (Livermore et al., 1990) and THA stability (Sikes van et al., 2008). Table 1 presents linear penetration rate observed for 28 mm and 32 mm diameter femoral heads. Only studies with an acetabular cup made of conventional UHMWPE and CoCr femoral head were taken to exclude the effect of material on wear. The range of linear penetration between the studies varies considerably indicating factors other than femoral head size influencing polyethylene wear.

Computational modeling aims to explain the implant wear from fundamental properties of UHMWPE (Liu et al., 2013). First studies have implemented Archard wear law in the finite element method simulations (Maxian et al., 1996) and in elasticity analysis (Pietrabissa et al., 1998). Further advances include incorporation of multidirectional motion effect on polyethylene wear (Kang et al., 2009), the effect of the contact pressure on the wear (Kang et al., 2009), elasto/plastic (Teoh et al., 2002), and time-varying visco-dynamic behaviour of UHMWPE (Quinci et al., 2014). In addition to Archard law, the contact area dependent (Liu et al., 2011) and the dissipation energy wear law (O'Brien et al., 2014) have been formulated.

Košak et al. (2011) proved that patient-specific hip contact stress is related to in vivo wear of total hip replacement. The contact stress in total hip replacement is influenced by operative parameters like lateral inclination of the acetabular cup (Hua et al., 2012) and patient-related parameters like magnitude and spectrum of loading forces (Liu et al., 2013; Takenaga et al., 2013). Although, the role of femoral head diameter and acetabular cup angle has been addressed in several mathematical models (Hua et al., 2012; Monif, 2012) and clinical studies (Little et al., 2009; Patil et al., 2003), the contribution of patient-specific geometry to polyethylene wear remains unclear. The aim of this study is to test

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Table 1

Review of linear penetration rates in THA studies. All studies included conventional UHMWPE as cup material and CoCr alloy as head material.

Author	Head diameter [mm]	Number of THA [1]	Follow-up [year]	Linear wear rate [mm/year]
Bankston et al. (1993)	28	162	6–8	0.05
Kabo et al. (1993)	28	23	5–19	0.23
Bankston et al. (1995)	28	77	6–8	0.05
Nashed et al. (1995)	28	74	5–10	0.17
Anseth et al. (2010)	28	25	15–20	0.15
Calvert et al. (2009)	28	60	0–4	0.13
Pedersen et al. (2001)	28	46	4–6	0.17
Engh et al. (2012)	28	88	9–13	0.11
Martell et al. (2003)	28	22	2–3	0.20
Hamilton et al. (2005)	28	30	3–11	0.11
Geerdink et al. (2006)	28	54	3–6	0.12
Nikolaou et al. (2012)	28	36	5	0.15
Hendrich et al. (2003)	28	109	2–13	0.14
Meftah et al. (2013)	28	31	15–20	0.14
Kabo et al. (1993)	32	9	6–16	0.21
Thanner et al. (1999)	32	76	7	0.13
Bono et al. (1994)	32	72	2–6	0.18
Kim et al. (1999)	32	52	10–11	0.29
Elfick et al. (1998)	32	47	0–12	0.23
Dowd et al. (2000)	32	47	4–11	0.18
Lee et al. (1999)	32	110	2–5	0.20

the hypothesis that interindividual variations due to patient's geometry of musculoskeletal system and placement of acetabular liner can have larger effect on wear than the radius of femoral head itself. The study is intended to further quantify contribution of selected patient-specific, surgical and design parameters on wear using a mathematical modeling in a study of patients with 28 mm and 32 mm diameter of the femoral head.

2. Methods

Implant position, geometry, material and both its kinematics and dynamics during daily loading cycles should be known to determine patient-specific THA wear. While implant type and position is usually known for given patient, kinematic and dynamic data on joint replacement are not available in orthopedic archives. In the following, the mathematical model of wear is derived that separates the effects of kinematics and dynamics on joint wear. The joint kinematics is assumed to be unified in all patients while the hip load is adopted for each patient by considering patient-specific hip loading in one-legged stance as representative loading condition for walking.

2.1. Hip wear model

According to Archard/Lancaster law, the linear wear of polyethylene depends on the wear coefficient k , contact stress p and the sliding speed v between the acetabular cup a femoral head (Maxian et al., 1996).

$$w = \int kp(t)v(t)dt \quad (1)$$

where the integration is performed over the loading period. Wear coefficient k depends on the contact stress and the cross-shear ratio (Kang et al., 2009).

$$k = e^a e^{b \ln Co} e^{c p_{AVE}} \quad (2)$$

where a, b, c are the constants, Co is the cross-shear ratio and p_{AVE} is the average stress at a given point during the loading cycle.

Let us consider a patient with a standard walking cycle. The cross-shear ratio at a given point of the acetabular cup is constant for a given motion cycle (Kang et al., 2009) while the sliding speed v is a function of femoral head radius $v = \omega r$, where ω is the angular speed at a given point. Furthermore, we can assume that variations in the contact stress during the motion are proportional to the average stress ($p(t) = p_{AVE} f(t)$), where $f(t)$ is a function describing time course of stress at given point with respect to the average stress. Using Eqs. (1) and (2), it can be written

$$w = r p_{AVE} e^{c p_{AVE}} e^a e^{b \ln Co} \int f(t) \omega(t) dt \quad (3)$$

It was shown that the hip contact stress in a one-legged stance p is proportional to the average stress during the walking cycle p_{AVE} (Daniel et al., 2008).

$$p_{AVE} = mp \quad (4)$$

where m is the proportionality constant. It was further shown that the linear penetration (d) position coincides to the maximum stress (p_{max}) position at the acetabular cup in a one-legged stance (Kořak et al., 2011). Using Eq. (4) in Eq. (3), the linear penetration d can be expressed as:

$$d = C I_{LW} \quad (5)$$

where

$$C = e^{a+b \ln Co} d N \int_{step} f(t) \omega(t) dt \quad (6)$$

accounts for kinematical effects like variations in sliding speed during the motion (Eq. (3)) or the cross-shear effect (Eq. (2)). We define I_{LW}

$$I_{LW} = p_{max} r e^{c m p_{max}} \quad (7)$$

as the wear index accounting for patient-specific acetabular cup load. The wear index I_{LW} is conditioned by intrinsic musculoskeletal geometry of the pelvis and proximal femur, diameter of femoral head as design parameter and orientation of acetabular cup as surgical parameter. Integration in Eq. (6) is performed over one loading cycle, N is the number of loading cycles in Eq. (6), and p_{max} in Eq. (7) is the peak contact stress in a one-legged stance.

2.2. Patients

Patients enrolled in this study were implanted with a cemented total hip replacement with cobalt chromium head and polyethylene cup (UHMWPE). Two groups of patients were involved: one group with 28 mm diameter of femoral head (6 male, 28 female, average age at implantation 67 ± 6.6 years) and another group with 32 mm diameter of femoral head (6 male, 23 female, average age 68 ± 4.1 years).

The contact stress in a one-legged stance was determined by the method described in Kořak et al. (2011). The method is based on biomechanical analysis of standard anteroposterior radiogram of pelvis and proximal femur and consists of calculation of hip joint reaction force and distribution of the force over the joint contact area.

The assessment of the hip joint reaction force \mathbf{R} is based on the force and the torque equilibrium of the body in a one-legged stance (Iglić et al., 2002) while the muscle forces are determined using inverse static optimization (Iglić et al., 2002) using minimization of muscle stress cubed as cost function (Crownshield and Brand, 1981). Hip joint force \mathbf{R} , radius of the acetabular cup r and abduction angle of the acetabular cup α (Fig. 1) are used to calculate THA contact stress. No clearance between the femoral head and the acetabular cup is assumed. Contact stress distribution was assumed to have cosine distribution and the peak contact stress (p_{max}) was determined by considering distribution of the total hip force over the weight-bearing area (Kralj-Iglić et al., 2012).

The model was adapted for each patient by scaling a generic musculoskeletal model of the hip on the basis of anteroposterior radiogram. Archive radiographs taken within one-week after implantation were used. The following geometrical parameters were measured from an anteroposterior radiogram (Fig. 1): height of the pelvis H , width of the pelvis C , vertical and horizontal position of the greater trochanter x and z , respectively and abduction angle of the acetabular cup α . The

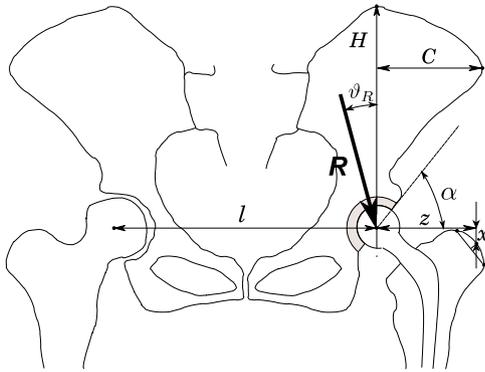


Fig. 1. Parameters used in mathematical model to assess the wear.

joint replacement type except head radius r is not included in the simulation and patient data serves to estimate variabilities in patient-specific geometry in both groups. All wear and stress distribution calculations were performed using custom scripts in MATLAB R2015b (The MathWorks, Natick, Massachusetts, USA).

2.3. Variance analysis

To quantify the effect of variability in a selected parameter, the given parameter could be taken as fixed in the wear model, e.g. individual inclination of the acetabular cup α is replaced by an average value of the group. Three groups of effects were studied: patient’s parameters, surgical parameters and design parameters. Patient’s parameters express individual geometry of the musculoskeletal system (Fig. 1) that affects the hip joint force \mathbf{R} . Surgical parameters are represented by the inclination of the acetabular cup α . Design parameters account for various femoral head diameter r . The effect of r on wear can be easily separated when considering hips with 28 mm and 32 mm femoral heads implanted.

2.4. Statistical analysis

Normality of variables distribution is tested by the Shapiro–Wilk test. The difference in all variables between the group of 28 and 32 mm femoral head is compared using an unpaired two-sided t -test. The homogeneity of variances between the groups is tested using the Fligner–Killeen test. Statistical analysis was performed using R Statistical Software (Foundation for Statistical Computing, Vienna, Austria). A p value of < 0.05 was considered significant. A box and whiskers plot was used in the variability study, the box defines the interquartile range, the line joining each box represents the median, and the whiskers represent the extreme data points.

2.5. Constants

The value of integral Eq. (6) is approximated in accordance with ISO 14242-1, where the angular speed for hip joint during the walking cycle is approximately 2 rad/s (Mattei et al., 2011). The number of steps per day is approximately 10^6 (Bohannon, 2007). The coefficients a, b, c, m and C_0 in Eq. (2) were determined as $-13.1, 0.19, -0.29, 0.48, 0.15$ respectively (Daniel et al., 2008; Kang et al., 2009). Using these values, the constant C for annual linear wear is $5.0 \text{ mm}^3\text{N}^{-1} \text{ m}^{-1}$. The weight of a standard patient is taken as 700 N.

3. Results

Musculoskeletal geometry of pelvis and proximal femur, except for the horizontal position of the greater trochanter z , showed no statistically significant differences between the group of patient with 28 and 32 diameter of implanted femoral heads. (Table 2). Smaller size of the femoral head increases the average contact stress in the group with 28 mm femoral head significantly. Predicted linear annual wear (d in Eq. (5)) is within the range observed in clinical studies (Table 1).

The range of feasible linear annual wear was obtained by including interindividual variations in musculoskeletal geometry and acetabular liner inclination (Table 2). Interindividual variations are more pronounced in hips with larger femoral head. The feasible linear annual wear shows that the effect of variations

between the patients is much higher than the effect of head diameter (Fig. 2).

Steeper acetabula have on average higher constant stress and wear (Figs. 3 and 4). The effect is more pronounced in 32 mm heads (Fig. 4). Data scattering depicts the effect of interindividual geometry.

By setting the cup angle α constant, the variance in wear is not significantly different from the original group, while setting identical geometry of pelvis and proximal femur significantly decreases variance in estimated linear annual wear (Table 3 and Fig. 5).

Table 2

Statistical difference in selected anatomical and biomechanical parameters between the group of patients with implanted femoral head of diameter 28 mm and 32 mm. Annual linear penetration $d = Cl_{LW}$ was estimated on the basis of mathematical model. $d = Cl_{LW}$.

Parameter	Units	28 mm	32 mm	p
l	[mm]	177.97 ± 11.86	181.28 ± 13.01	0.296
x	[mm]	16.58 ± 8.83	14.56 ± 10.06	0.398
z	[mm]	54.69 ± 6.79	61.26 ± 10.02	0.003*
C	[mm]	53.91 ± 8.41	53.98 ± 9.68	0.973
H	[mm]	118.39 ± 14.15	124.48 ± 16.51	0.120
α	[°]	42.24 ± 8.56	45.48 ± 8.51	0.146
R	[kN]	1.96 ± 0.19	1.86 ± 0.20	0.036*
θ_R	[°]	8.84 ± 2.21	9.83 ± 2.26	0.085
p_{max}	[MPa]	5.31 ± 0.902	3.89 ± 0.597	$< 0.001^*$
Cl_{LW}	[mm]	0.175 ± 0.006	0.179 ± 0.011	0.080

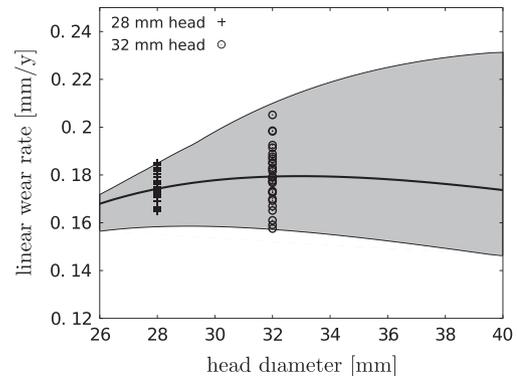


Fig. 2. Estimated linear wear rate for THA with different diameters of femoral head. The solid curve corresponds to a patient with average geometry of the pelvis and proximal femur. The shaded area corresponds to feasible linear wear rate computed from data of all patients.

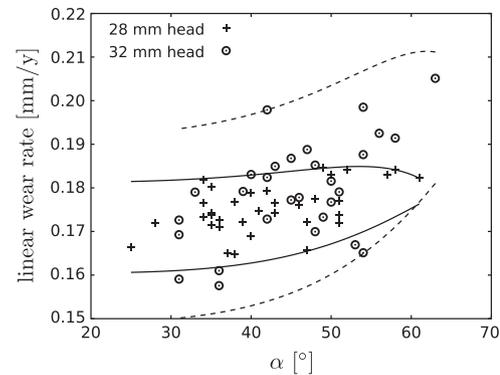


Fig. 3. Peak contact stress in a one-legged stance as a function of inclination of acetabular cup (α) for THA with various diameters of femoral head. Solid and dashed lines correspond to theoretically extreme values for 28 and 32 mm femoral head diameter, respectively.

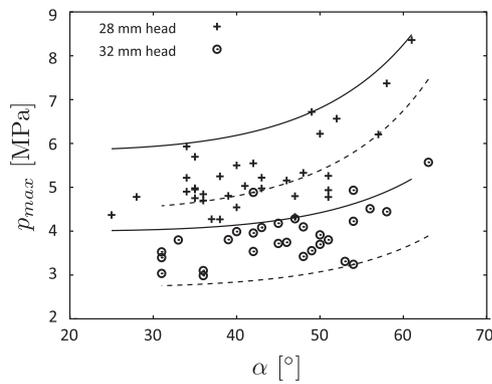


Fig. 4. Estimated annual wear $d = Cl_{LW}$ as a function of inclination of acetabular cup (α) for THA with various diameters of femoral head. Solid and dashed lines correspond to theoretically extreme values for 28 and 32 mm femoral head diameter, respectively.

Table 3

p - values obtained using the Fligner–Killeen test of homogeneity of variances for null hypothesis that the variances in estimated hip contact stress and linear wear in between the original and the modified group are the same.

	28 mm		32 mm	
	α fixed	R fixed	α fixed	R fixed
p (stress variance)	0.348	0.019*	0.356	0.024*
p (wear variance)	0.630	0.002*	0.413	0.022*

4. Discussion

This study is intended to test the hypothesis that inter-individual variations due to patient's hip geometry and acetabular cup angle have larger effect on wear than the radius of the femoral head itself. Mathematical modeling is used to quantify the contribution of selected patient-specific, surgical and design parameters to linear wear in a study of patients with 28 mm and 32 mm implanted diameter of the femoral head.

Results in this study explain the same level of annual linear wear observed for THAs with 28 and 32 mm femoral head in clinical studies (Table 1) as the linear wear is shown to be insensitive to femoral head diameter (Fig. 2). The dependence between the wear and head diameter described in Fig. 2 is in agreement with the finite element simulations provided by Kang et al. (2009). Presented study shows that in some cases, the 32 mm heads resulted in even lower wear than the 28 mm heads as was also observed in clinical studies (Table 1). Large scattering of data in Fig. 2 and from clinical studies be attributed to the variability between the patients (Table 3) and possible variability in methods of wear estimation. Cup angle is also a major consideration affecting THA wear besides the radius of femoral head. The conventional 'safe zone' for acetabulum inclination is taken between 30° to 50° (Pedersen et al., 2005) that is a compromise between containment and impingement (Widmer, 2007). Models reported by D'Lima et al. (2001), Oki et al. (2004), and Patil et al. (2003) found larger contact stresses at more vertically oriented cups and concluded that cup abduction angle influences polyethylene wear in THA. These findings were further supported by experimental and clinical studies showing higher linear wear in hips with acetabular angle more than 45° (Little et al., 2009; Patil et al., 2003; Wan et al., 2008). In contrast, finite element models of Hua et al. (2012) and Korhonen et al. (2005) show that the contact stresses might be insensitive to the cup inclination as was also observed at two year follow-up study (Kadar et al., 2012). Presented analysis states that both of these mechanisms are

feasible, depending on the individual hip geometry. If the contact stress for a given patient is high for a given cup angle, increase in cup angle α yields considerable increase in peak contact stress (Fig. 3) as also observed by Patil et al. (2003). This effect is known to be related to hip joint reaction force pointing closer to the acetabular rim (Kralj-Iglić et al., 2012) that might occur in dysplastic patients studied by Oki et al. (2004). If the contact stress is relatively low at a given cup angle, i.e. hip joint force acts more towards the center of the cup (Kralj-Iglić et al., 2012), the effect of cup angle on contact stress is much lower (Fig. 3). Small influence of cup inclination on contact stress observed by Hua et al. (2012) can be explained by a larger than average inclination of hip joint force being used (Table 2).

The estimated linear annual wear increases with cup angle in general that is in agreement with Patil et al. (2003) and Oki et al. (2004). Our study shows that the effect of patient's specific musculoskeletal geometry is more pronounced on wear than on contact stresses. For example, we identify hips with identical cup angle of 54° but different predicted wear rate of 0.165 mm/y and 0.20 mm/y. The proper orientation of acetabular cup is more important in hips with 32 mm heads where the acetabular cup angle higher than 50° in combination with unfavorable patient's musculoskeletal geometry could result in considerable increase in THA wear (Fig. 4). The study indicates that 28 mm head should be considered in the case of steep acetabula as it is less susceptible to wear (Fig. 4).

The main advantage of mathematical modeling is in its predictive abilities (Table 3 and Fig. 5). In addition to the proper acetabular cup position (Patil et al., 2003); variance analysis indicates that patient-specific parameters should be considered to reduce the wear. For example, coxa valga or excessive femoral anteversion decreases abductor lever arm that results in a biomechanically unfavourably high hip loading force pointing closer to the cup rim. Smaller cup angle should be considered in these patients to reduce the risk of higher wear (Fig. 4). The results are in agreement with other studies showing importance of patient-specific geometry in estimating hip joint loading force (Correa et al., 2011), biomechanical status of dysplastic hips (Mavčič et al., 2004) or hips after aseptic necrosis of femoral head (Daniel et al., 2006).

Linear penetration d is the principal variable considered in the presented study as it is the primary variable reported in multiple clinical studies (Table 1). The number of wear particles created is determined by volumetric wear that is directly related to the square of the radius of the head (Kořak et al., 2011). Radius of femoral head should further be taken into account if considering wear rate (Atkins et al., 2011).

The model adopts several assumptions. The prediction of linear wear is based on average patient constants related to motion kinematics and unified level walking activity. The interpersonal variation in motion activities, kinematics and dynamics of walking and weight of the patients could further increase the variation in the hip joint reaction force (Takenaga et al., 2013). The estimation of hip joint reaction force variation within this study is based on intrinsic geometry of patient's hip only and it could therefore be considered as conservative estimation. The interpersonal variations mentioned above would also affect the wear by affecting sliding velocity and cross-shear ratio (Eq. 3). Furthermore, the number of annual walking cycles decreases with patient's age (Bohannon, 2007). It implicates higher wear in younger patients as observed by Kim et al., 1999. Larger variation in hip joint reaction force and wear parameters would contribute to even larger scattering in wear data in Fig. 4 and further decrease the effect of femoral head radius on wear in Fig. 5.

The cup impingement and edge contact (Klues et al., 2007), known to increase the contact stress, are not directly included

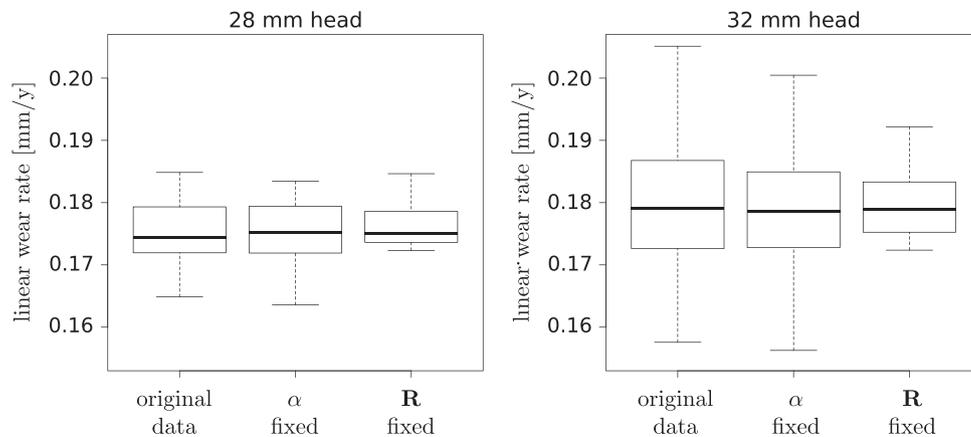


Fig. 5. Boxplot of estimated linear wear for patients who have undergone THA with 28 mm and 32 mm femoral heads. The estimated linear wear for the patients was computed by omitting variation in selected input variables.

within the model. These effects contribute additional mechanisms to the studied adhesion wear at bearing surface. The model predictions are therefore limited in acetabular component with steep inclination and excessive anteversion that contribute to edge loading (Hua et al., 2014). The wear might further alter contact surfaces, cause the misalignment of the centres of the head and the cup, and influence hip contact stress distribution (Wang, 2015). Although, finite element simulation shows that femoral head penetration does not considerably affect initial contact mechanics (Hua et al., 2012).

Conventional polyethylene was adopted in this study, as conventional polyethylene THA components have been followed-up in many clinical and experimental studies that provide the theoretical basis of this presented study. New cross-linked polyethylenes are presumed to give lower wear rates (Nikolaou et al., 2012) and are more used in current arthroplasty. The mathematical model developed within this study could be adopted for cross-linked polyethylene by modifying wear coefficients that would result in lower estimated linear wear and lower variance in wear magnitude (Abdelgaied et al., 2012). The principal wear mechanisms in conventional and crosslinked polyethylene are the same and we assume that the general conclusions of the study in quantification of patient-specific and surgical parameters are valid for cross-linked polyethylene as well, although the absolute value of wear would be lower. This assumption is supported by the study of Lachiewicz et al. (2009) that shows no association between the femoral head diameter and the linear wear rate in acetabular cups made from crosslinked polyethylene and study of Allepuz et al. (2014) showing that the risk of revision did not differ significantly between < 28-mm and 32-mm head sizes articulating with highly cross-linked polyethylene cup.

The study provides an insight into the importance of individual factors influencing THA wear that should be considered in pre-operative planning. The study shows that the acetabular cup orientation should be planned by considering patient specific geometry, for hips with 32 mm heads and steep acetabula particularly. Further research is warranted to provide definite relationship between the patient's musculoskeletal geometry, the specific motion pattern, and the optimal cup orientation.

Estimation of linear wear rate indicates that polyethylene linear wear is almost insensitive to the change of diameter of femoral head. Predicted annual linear wear depends mostly on abduction angle of the acetabular liner and individual geometry of the musculoskeletal system of the hip. The variation of patient-specific hip geometry was shown to have the greatest effect on variation in linear wear rate.

Conflict of interest

None.

Acknowledgments

This research is supported by AZV MZ CR project No. 15-31269A.

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