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Association between contact hip stress and RSA-measured wear rates in total hip arthroplasties of 31 patients

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Abstract

Background: The main concern in the long run of total hip replacements is aseptic loosening of the prosthesis. Optimization of the biomechanics of the hip joint is necessary for optimization of long-term success. A widely implementable tool to predict biomechanical consequences of preoperatively planned reconstructions still has to be developed. A potentially useful model to this purpose has been developed previously.

The aim of this study is to quantify the association between the estimated hip joint contact force by this biomechanical model and RSA-measured wear rates in a clinical setting.

Methods: Thirty-one patients with a total hip replacement were measured with RSA, the gold standard for clinical wear measurements. The reference examination was done within 1 week of the operation and the follow-up examinations were done at 1, 2 and 5 years. Conventional pelvic X-rays were taken on the same day.

The contact stress distribution in the hip joint was determined by the computer program HIPSTRESS. The procedure for the determination of the hip joint contact stress distribution is based on the mathematical model of the resultant hip force in the one-legged stance and the mathematical model of the contact stress distribution. The model for the force requires as input data, several geometrical parameters of the hip and the body weight, while the model for stress requires as input data, the magnitude and direction of the resultant hip force. The stress distribution is presented by the peak stress—the maximal value of stress on the weight-bearing area (p_{\max}) and also by the peak stress calculated with respect to the body weight (p_{\max}/W_B) which gives the effect of hip geometry.

Visualization of the relations between predicted values by the model and the wear at different points in the follow-up was done using scatterplots. Correlations were expressed as Pearson r values.

Results: The predicted p_{\max} and wear were clearly correlated in the first year post-operatively ($r = 0.58$, $p = 0.002$), while this correlation is weaker after 2 years ($r = 0.19$, $p = 0.337$) and 5 years ($r = 0.24$, $p = 0.235$). The wear values at 1, 2 and 5 years post-operatively correlate with each other in the way that is expected considering the wear velocity curve of the whole group.

The correlation between the predicted p_{\max} values of two observers who were blinded for each other's results was very good ($r = 0.93$, $p < 0.001$).

Conclusion: We conclude that the biomechanical model used in this paper provides a scientific foundation for the development of a new way of constructing preoperative biomechanical plans for total hip replacements.

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Keywords: Hip joint contact forces; Wear; RSA; Total hip arthroplasty

1. Introduction

The total hip arthroplasty may be considered as the most successful joint replacement in orthopedic surgery. The

main concern in the long run is aseptic loosening of the prosthesis. The exact mechanism causing aseptic loosening is still unknown. Currently, the most accepted view is that the effects of polyethylene wear particles are the most important cause of failure (Santavirta et al., 1990; Willert et al., 1990; Cooper et al., 1992). Another suggested mechanism of failure states that early (undetected) mechanical

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loss of fixation causes periprosthetic granulomas and excessive wear (through a pathway of micromotion with hydrostatic pressure effects and subsequent bone resorption, thus potentially leading to a self-perpetuating process), thus reversing the cause and effect of the previously mentioned theory (Mjoberg, 1994). Animal experimental model studies provide support for this possibility (De Man et al., 2005; Skripitz and Aspenberg, 2000; van der Vis et al., 1998a, b). The truth might even be a combination of these pathways.

With either of these two plausible pathways of loosening, optimization of the biomechanics of the hip joint is necessary for optimization of chances of success; minimizing the hip joint contact force will either minimize wear rates of the cup, or it will minimize the probability of early mechanical loss of fixation of the prosthetic components. This has intuitively already long been accepted. Preoperative planning has always been based on the concepts of restoration of certain biomechanical parameters, like the center of rotation, femoral offset and leg length equalization (Schmalzried, 2005; Maloney and Keeney, 2004; Fessy et al., 1999; Legal and Ruder, 1979). Several studies support this approach by providing evidence of a relation between these parameters and hip joint contact forces (Kleemann et al., 2003; Carls et al., 2002; Lengsfeld et al., 2000; Doehring et al., 1996; McGrory et al., 1995; Vasavada et al., 1994; Karachalios et al., 1993; Davey et al., 1993; Johnston et al., 1979; O'Toole III et al., 1995). The surgeon should therefore preoperatively be informed on which reconstruction would lead to a biomechanically optimized reconstruction. Highly specialized three-dimensional applications do exist, but have no place in daily practice since their burden in time and costs are considered too heavy (Noble et al., 2003). A widely implementable tool which directly links the preoperatively planned reconstruction to a prediction of the resulting forces yet has to be developed.

The effect of different geometrical configurations of the pelvis and hip joint on the hip joint contact force have been investigated in earlier publications (Vengust et al., 2001; Ipavec et al., 1999; Kersnic et al., 1996; Iglıc et al., 1993, 1995; Srakar et al., 1992). A model which uses input from standard plain pelvic radiographs was clinically applied to assess pathologic conditions such as developmental dysplasia of the hip, outcome after intertrochanteric osteotomies, Chiari and Salter osteotomies, and surgical treatment for slipped capita femoris (Dolinar et al., 2003; Pompe et al., 2003; Herman et al., 2002; Zupanc et al., 2001). It would be valuable if such a biomechanical model could be implemented to enhance preoperative planning and post-operative prognostic assessment of total hip arthroplasties.

The aim of this study is to quantify the association between the estimated hip joint contact force by this biomechanical model and radiostereometric analysis (RSA)-measured wear rates in a clinical setting.

2. Materials and methods

RSA is a method with which change of position of bony or artificial (prosthetic) structures in relation to each other can be measured very accurately. Several tantalum balls need to be incorporated in the objects of interest for this measurement technique. A stereoradiographic approach provides a computer with the necessary data to calculate the relative displacement of interest.

In our study, 31 patients received a ScanHip system with Opticup (Biomet, UK). The cup was marked with 7–9 tantalum markers by the manufacturer, and was available in sizes of 46–56 mm. RSA examinations were done by a uniplanar technique (Karrholm et al., 1997; Selvik, 1989), with the patient in supine position. The two X-ray sources were fixed (mounted to the ceiling). We used a type-41 calibration cage (Tilly Medical, Sweden) and the computer software UmRSA version 5.0 (RSA Biomedical, Sweden). The reference examination was done within 1 week of the operation and the follow-up examinations were done at 1, 2 and 5 years. Conventional pelvic X-rays were made on the same day.

The contact stress distribution in the hip joint was determined by the HIPSTRESS computer program (Iglıc et al., 2002). The procedure for the determination of the hip joint contact stress distribution is based on the mathematical model of the resultant hip force in the one-legged stance (Ipavec et al., 1999) and the mathematical model of the contact stress distribution (Mavcic et al., 2002). The model for the force requires as input data the geometrical parameters of the hip: the inter-hip distance l , the pelvic height H and width C , and the point of insertion of the effective muscle (its vertical coordinate x and horizontal coordinate z with respect to the origin in the center of the articular sphere) (Fig. 1) and the body weight W_B , while the model for stress requires as input data the magnitude R and direction ϑ_R of the resultant hip force, the radius of the hip joint articular surface (r) and the centre-edge angle of Wiberg (ϑ_{CE}) (Fig. 1). The body weight of the patients was measured immediately before the operation. The geometrical parameters were determined from the standard anteroposterior (AP) plain pelvic X-rays using the QBone[®] Planner 5.4 software (Medis, Leiden, The Netherlands). Magnification was corrected for by calibration of each X-ray using the 28 mm prosthetic femoral head as the reference object. The stress distribution is presented by the peak stress—the maximal value of stress on the weight-bearing area (p_{max}) and also by the peak stress calculated with respect to the body weight (p_{max}/W_B) which gives the effect of hip geometry (Sychterz et al., 1999). Low p_{max} and p_{max}/W_B are biomechanically favorable, while high p_{max} and p_{max}/W_B are biomechanically unfavorable.

Visualization of the relations between predicted values by the model and the wear at different points in the follow-up was done using

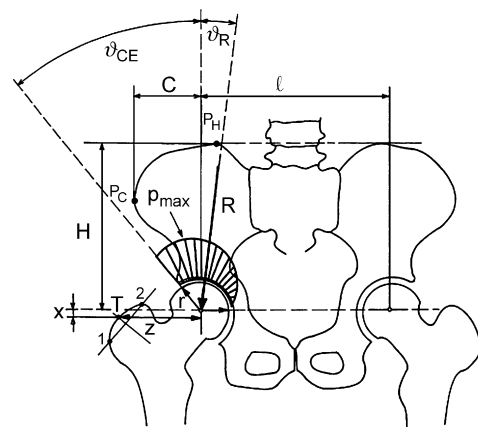


Fig. 1. The geometrical parameter of the hip and pelvis needed for determination of the maximal stress on the weight-bearing area. The stress distribution and the resultant hip joint force R are also shown schematically. Symbol ϑ_R denotes the inclination of R with respect to the vertical and the point T denotes the effective muscle attachment point on the greater trochanter.

scatterplots. Correlations were expressed as Pearson r values. All statistics were performed using SPSS version 12.0 (SPSS Inc., Chicago).

3. Results

The mean age of the patients was 67 years (range 51–80). Eighteen patients (58%) were female, and 19 patients (61%) were operated on their left hip. All patients were operated on for degenerative arthritis of the hip joint.

The predicted p_{\max} and wear were clearly correlated in the first year post-operatively ($r = 0.58$, $p = 0.002$), while this correlation is weaker after 2 years ($r = 0.19$, $p = 0.337$) and 5 years ($r = 0.24$, $p = 0.235$) (see Table 1). The wear values at 1, 2 and 5 years post-operatively correlate with each other in the way that is expected considering the wear velocity curve of the whole group (see Fig. 2).

The correlation between the predicted p_{\max} values of two observers who were blinded for each other's results was very good ($r = 0.93$, $p < 0.001$).

4. Discussion

The importance of biomechanically favorable hip joint replacements has been stressed by many authors, but no technique to aid the clinician in determining the most favorable preplanning of a total hip arthroplasty has been available yet. A validated three-dimensional biomechanical model of the hip joint was used for measuring the peak hip joint contact stress in a series of patients which were followed to obtain wear measurements by RSA (which is the gold standard for in vivo wear measurements) at 1, 2 and 5 years post-operatively.

The biomechanical model to determine the hip joint stress and force has been validated and used in several other research areas focusing on developmental dysplasia of the hip, outcome after intertrochanteric osteotomies, Chiari and Salter osteotomies, and surgical treatment for slipped capita femoris (Dolinar et al., 2003; Pompe et al., 2003; Herman et al., 2002; Zupanc et al., 2001). It is the only existing model that yields stress and force estimations from a three-dimensional model, but requires only data which can be retrieved from plain pelvic X-rays.

The calculated contact stress distribution in the hip joint is based on a mathematical model of the resultant hip force in one-legged stance and a mathematical model of the radial contact stress distribution. The mathematical model

used for calculation of contact stress distribution is based on the assumption that the radial stress on the articular surface of the hip is proportional to the cosine stress distribution function. The above assumption was originally adopted only for intact hips (Ipavec et al., 1999; Brinckmann et al., 1981), where the cosine stress distribution function originates in the deformation of the cartilage layer due to a small displacement of the femoral head with respect to acetabulum as a consequence of the load transmitted through the hip joint (described by the resultant hip force R). Later, it was shown that the above-described cosine radial stress distribution for intact hips can also be used to describe the radial (normal) stress and the linear wear in artificial hips where the cosine stress distribution originates in protrusion of the head of the femoral part of the prosthesis into the polyethylene acetabular liner of the prosthesis (Kosak et al., 2003).

It should be noted that the mathematical model predicts that the maximum stress can certainly be found on the cranialateral side of the acetabular component, despite the

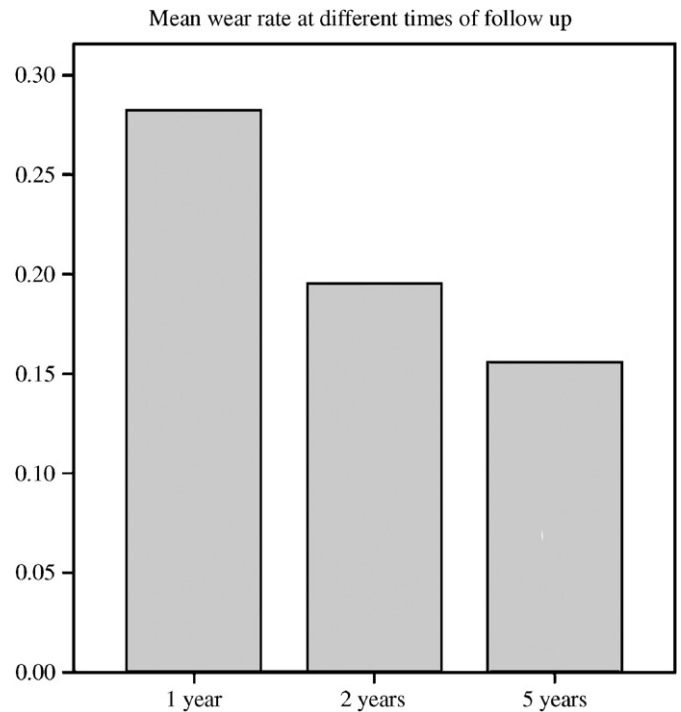


Fig. 2. The mean wear rate (mm/year) at three moments of follow-up: 1, 2 and 5 years post-operative.

Table 1
Correlation matrix of p_{\max} and post-operative wear

	p_{\max}	1 year	2 years	5 years
p_{\max}		0.53 ($p = 0.005$)	0.23 ($p = 0.241$)	0.22 ($p = 0.274$)
1 year	0.53 ($p = 0.005$)		0.46 ($p = 0.020$)	0.42 ($p = 0.038$)
2 years	0.23 ($p = 0.241$)	0.46 ($p = 0.020$)		0.84 ($p < 0.001$)
5 years	0.22 ($p = 0.274$)	0.42 ($p = 0.038$)	0.84 ($p < 0.001$)	

Note: All correlations are expressed as Pearson correlation coefficients. p_{\max} is the peak stress as predicted by the HIPSTRESS algorithm, and 1, 2 and 5 years refer to the amount of wear (mm) at each of those follow-up moments.

craniomedial direction of the hip joint contact force. This is an interesting finding which should be taken into account when discussing the mechanisms behind cranio-lateral wear (Murray and O'Connor, 1998).

The normalized peak contact stress and the size of the weight-bearing area have been estimated from the anteroposterior radiographs already before (Brinckmann et al., 1981). However, in this study the hip joint contact force was determined by using a simple mathematical model, which considers only a single effective muscle group (the hip abductors) and very few direct musculoskeletal anatomical data. Consequently, the variation of the model parameters by means of the variation of the geometrical parameters of the femoral and pelvic shape were strongly limited. More accurate estimation of the hip joint contact force requires a three-dimensional muscle model. In this study, the hip joint contact force is calculated by using a three-dimensional mathematical model that includes nine effective muscles (Igljic et al., 2002). The coordinates of the muscle pelvic and femoral attachment points are taken from another study (Dostal and Andrews, 1981). As a consequence, the predicted values of the average muscle tensions and the magnitude of the hip joint resultant force (R) can be determined, taking the individual variations in femoral and pelvic configuration into account. All the rescaling is effectuated in the three-dimensional model which directly determines the outcome, but solely using input which can be derived from a plain pelvic radiograph.

The hip joint contact force and stress are dependent on more parameters than can be measured from a plain pelvic X-ray. It is clear that a patient who varies his activity pattern will change his wear rate, while the predicted stress load of his hip joint will remain constant. This factor is not accounted for in this model and might explain most of the variance in the data. The fact that the correlation between predicted p_{\max} and measured wear is never higher than 0.6 is thus a reflection of the heterogeneity of the patient population. High correlations can only be expected if the chosen study population is strictly selected, like a group of patients of similar age, in which all had aseptic loosening of their prosthesis within 10 years. In any case, for preplanning of a particular patient this is not an issue, since the main concern is which reconstruction would provide an optimized situation for that individual. The fact that this individual patient might influence the absolute stress load over time by being more or less active, does not change the fact that he has been given an optimized reconstruction.

Our analysis pointed out that body weight as a separate variable seemed to be correlated with the measured wear, but it was outperformed by the p_{\max} variable. Using multivariate analysis with wear as the independent variable, the only remaining unconfounded covariate was also p_{\max} . However, p_{\max} is an interaction variable (representing effect modification) integrating body weight and p_{\max}/W_B . In other words, body weight can be considered an effect modifier on the relation between biomechanical geometry (or predicted stress) and wear, or

the other way round—biomechanical geometry (or predicted stress) can be considered an effect modifier for the relation between body weight and wear.

The magnitudes of the correlations and their change over time are roughly as expected. The correlation between the predicted p_{\max} and wear is the strongest in the first post-operative year, because the rehabilitation period results in relatively similar activity patterns; patients get similar instructions and (para)medical support in the first post-operative period which will diminish inter-patient differences. After the first year, these inter-patient differences will certainly be a source of variance, which results in a decrease of the correlation between the predicted p_{\max} and measured wear after 2 and 5 years. A second possible explanation is the fact that in the first year(s) the predominant reason for penetration of the femoral head into the liner is creep, which is expected to be strongly dependent on p_{\max} , while wear in the following years is influenced by other factors as stated above. Complicating factors such as third body wear are also expected to be more prominent as time progresses.

The wear values at 2 and 5 years post-operatively are strongly correlated with each other, while wear after 1 year is not as strongly correlated with wear at 2 or 5 years. This is a direct consequence of bedding-in and creep which predominantly occur in the first post-operative year (45).

We conclude that the biomechanical model used in this paper provides a scientific foundation for the development of a new way of constructing preoperative biomechanical plans for total hip replacements. Future investigations might focus on confirming the relations we describe, optimization of the translation of a biomechanical preoperative plan into the actual reconstruction, and survival analysis of large cohorts.

Conflict of interest

All the authors declare that none of them has proprietary, financial, professional or other personal interests of any nature or kind that could be construed as influencing the position presented in this paper.

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