Backside volumetric change in the polyethylene of uncemented acetabular components

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Polyethylene wear of acetabular components is a key factor in the development of periprosthetic osteolysis and wear at the articular surface has been well documented and quantified, but fewer data are available about changes which occur at the backside of the liner.

At revision surgery for loosening of the femoral component we retrieved 35 conventional modular acetabular liners of the same design. Linear and volumetric articular wear, backside volumetric change and the volume of the screw-head indentations were quantified. These volumes, clinical data and the results from radiological Ein Bild Röntgen Analyse migration analysis were used to identify potential factors influencing the volumetric articular wear and backside volumetric change.

The rate of backside volumetric change was found to be 2.8% of the rate of volumetric articular wear and decreased with increasing liner size. Migrated acetabular components showed significantly higher rates of backside volumetric change plus screw-head indentations than those without migration.

The backside volumetric change was at least ten times larger than finite-element simulation had suggested. In a stable acetabular component with well-anchored screws, the amount of backside wear should not cause clinical problems. Impingement of the screw-heads could produce more wear particles than those generated at the liner-shell interface. Because the rate of backside volumetric change is only 2.8% of the rate of volumetric articular wear and since creep is likely to contribute a significant portion to this, the debris generated by wear at the backside of the liner may not be sufficient to create a strong osteolytic response.

Periprosthetic osteolysis and subsequent loosening of the components are major complications of hip replacement.1 The relationship between periprosthetic osteolysis and wear debris from polyethylene has been well established2-8 as has polyethylene wear of the articulating surface between the head and the acetabular component.2,9-11

The interface between the liner and the metal backing is also a source of polyethylene particles.12-18 Micro-movement at this interface can wear away the machining and manufacturer’s markings and also lead to cold flow of polyethylene into screw-holes or screw-heads. The micromovement seen in the first generation of modular acetabular components was due to poor locking mechanisms.19-21

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Terms such as backside wear, back surface wear or backside deformation are used to describe this phenomenon.13,14 We prefer the term backside volumetric change since it includes creep and wear. Volumetric articular wear may also have a creep component which appears not to be a significant proportion of the overall change at the articular surface.11

The role of backside wear in modular hip prostheses is controversial13,14,22 and many aspects have been investigated.9,12,20,23 Visual assessment of the back of the polyethylene liner13,17,22,23 and the measurement of micromovement between it and the metal backing has been performed.14,19-23 In vitro analyses have investigated different designs, the influence of features such as the type of locking mechanism, the number of screw-holes, the use of screws, the size of the liner and its composition and the surface finish.12,23,24 Kurtz et al16 calculated volumetric articular wear and backside volumetric change rates using a finite-element model and concluded that the latter were at least three orders of magnitude less than frontside rates. In order to improve the liner-backing interface, the conformity of the shell and liner was increased, the inner surface
of the liner was smoothed and non-modular, pre-assembled, cementless acetabular components were developed.5,17,25,26

To our knowledge there are no previous reports which have directly measured and compared the rates of volumetric articular wear and of backside volumetric change. Our study was undertaken to determine the extent of both, the contribution of screw-head indentations to backside changes, the role of factors such as gender, age, weight and height, and the size or inclination angle of the liner and whether there was a correlation between changes at the liner metal interface and migration of the acetabular component.

Patients and Methods
We retrieved 35 polyethylene liners (single-design SL cup; ME Müller, Zimmer GmbH, Winterthur, Switzerland) with a median implantation time of 54 months (49 to 66). All the patients were undergoing revision total hip replacement for loosening of the femoral component. This exceptional frequency of femoral revision was related to the use of cemented titanium stems.27 The acetabular components had all been implanted in one hospital using the same operating technique. The shell was of pure titanium according to ISO 5832-2, with a fine-blasted inner surface (roughness value 1 μm to 2 μm), five screw-holes and three slots (Fig. 1) and all were fixed with three to five screws. The polyethylene kGy liners were machined from pressed sheets of GUR 1020 and gamma sterilised under nitrogen at a dose of 25 kGy to 40 kGy. The liner was locked into the shell by a snap-fit mechanism with matching toothed rims for rotational stability. In order to avoid deformation at the articular and backside surfaces the liners were removed with a special tool inserted into four holes in the rim of the liner. After the insertion of the tool the liner was levered out with a chisel under the polyethylene tooth. There were no signs of third-body wear on the articular aspect or backside of the liner.

The liners were 52 mm, 56 mm or 60 mm in diameter and were all paired with a 28 mm Biolox (Ceramtec, Plochingen, Germany) femoral head made of alumina ceramic according to ISO 6474.

During revision, the surgeon checked the stability of the screws by attempting to tighten them. Any which could be turned were considered to be loose and this occurred in 12 of the 35 acetabular components. Four were unstable after removal of the loose screws. The remaining 31 were from stable components.

Baseline details and clinical data were derived from the hip registry of the Maurice E Müller Institute for Evaluative Research in Orthopaedic Surgery at the University of Berne, Switzerland (Tables I and II).

The measurement of total linear articular wear was performed according to the cast technique of Buchhorn et al.,28 whereby a cast of the retrieved liner was compared with that of a new one. Both retrieved and new liners had the same manufacturing tolerances. The total linear articular wear was defined as the maximum difference between the original and the measured radius which indicated the highest local wear. The volumetric articular wear was calculated from the total linear articular wear using a formula described by Dowson et al.29

The backside surface of the liner was sputter-coated with gold to ensure clearer demonstration of the polyethylene extrusions through the screw holes. Backside changes were evaluated using a stereomicroscope (Wild M3B; Wild Heerbrugg Ltd, Heerbrugg, Switzerland) with a magnification of up to ×40.

The creep of the polyethylene into the screw-holes or slots induced a difference in height between the extrusions and the surrounding surface. We used the term ‘imprint areas’ for the area formed between the extrusions. Using the stereomicroscope, points for height difference were localised and measured on a co-ordinate measuring machine (CMM5, SIP, Geneva, Switzerland). For easier surface calculation the imprint areas were divided into different rectangles (Fig. 2), the length and width of which were measured by a digital caliper. Since the height of the extrusions surrounding the rectangles was variable, measurements were taken at different points along the edges of the extrusions whereupon the three greatest heights surrounding a rectangle were used to determine a mean extrusion height. We used this and the surface area to calculate a volume and all partial volumes of a liner added up to a single volume. This was the backside volumetric change because it was considered to be an estimation of the volume resulting from changes on the backside of a liner as a result of the combination of creep and wear.

In order to show clearly the influence of screws on backside change, the volume of the screw-head indenta-
tions was not incorporated in the backside volumetric change but analysed separately. The depth of these indentations in the polyethylene was measured by a co-ordinate measuring machine and the other indentation dimensions by a digital caliper. The volume of the screw-head indentations was calculated from these values as an estimation of the true change in volume resulting from creep and wear. The shape of the screw-head indentations provided information on the position of the screw with respect to the liner, but this aspect was not examined in our study.

Migration of the acetabular component was assessed by one of the authors (AHK) using the Ein Bild Röntgen Analyse (EBRA) computerised method for the radiological assessment of migration.\textsuperscript{30-32} A minimum of four standard pelvic radiographs is required for measurements with an accuracy of < 1 mm of measuring error (95% confidence limits).\textsuperscript{30,32} Standardised pelvic radiographs, with the patient’s legs dangling, were obtained after four months and one, two and five years. Measurements of < 1 mm were considered to be ‘no migration’ (n = 23) and > 1 mm, ‘migration’ (n = 12).

\textbf{Statistical analysis.} The data were described as the medians and quartiles (IQR, interquartile range) for continuous endpoints. Binary endpoints were characterised by frequen-

\begin{table}[h]
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\caption{Details (median, IQR\textsuperscript{*}) of the volumetric articular wear (VAW) and backside volumetric change (BVC)}
\begin{tabular}{lcccc}
\hline
& \textbf{Linear articular wear rate (mm/year)} & \textbf{VAW rate (mm\textsuperscript{3}/year)} & \textbf{BVC rate (mm\textsuperscript{3}/year)} & \textbf{BVC rate + screw-head indentation (mm\textsuperscript{3}/year)} \\
\hline
\multicolumn{5}{l}{All liners (n = 35)} \\
& 0.08 (0.06 to 0.11) & 32.86 (17.75 to 45.61) & 0.69 (0.38 to 1.20) & 1.10 (0.63 to 2.76) \\
Male patients (n = 30) & 0.08 (0.06 to 0.10) & 31.98 (17.67 to 44.00) & 0.54 (0.35 to 1.13) & 1.11 (0.63 to 2.26) \\
Female patients (n = 5) & 0.08 (0.05 to 0.14) & 32.86 (13.29 to 62.01) & 1.10 (0.76 to 3.03) & 2.92 (0.76 to 4.81) \\
p-value & 0.766 & 0.699 & 0.054 & 0.219 \\
\hline
\multicolumn{5}{l}{Acetabular component diameter (mm)} \\
52 (n = 6) & 0.09 (0.07 to 0.14) & 36.75 (22.71 to 57.70) & 1.67 (0.98 to 2.80) & 2.79 (1.50 to 4.11) \\
56 (n = 20) & 0.09 (0.06 to 0.11) & 31.98 (19.24 to 48.66) & 0.70 (0.50 to 1.18) & 0.91 (0.66 to 2.35) \\
60 (n = 9) & 0.06 (0.04 to 0.10) & 20.15 (9.22 to 38.21) & 0.36 (0.18 to 0.45) & 1.11 (0.31 to 1.95) \\
p-value & 0.614 & 0.614 & 0.023 & 0.046 \\
52 vs 56 & 0.145 & 0.113 & 0.012 & 0.066 \\
56 vs 60 & 0.216 & 0.365 & 0.013 & 0.627 \\
\hline
\multicolumn{5}{l}{No migration (n = 23)} \\
& 0.08 (0.06 to 0.10) & 29.28 (17.42 to 39.09) & 0.50 (0.30 to 1.10) & 0.82 (0.49 to 1.70) \\
Migration (n = 12) & 0.09 (0.07 to 0.12) & 37.70 (22.50 to 51.20) & 0.95 (0.56 to 2.57) & 2.21 (1.18 to 3.13) \\
p-value & 0.310 & 0.263 & 0.073 & 0.011 \\
Stable acetabular components (n = 31) & 0.08 (0.06 to 0.11) & 30.23 (17.42 to 49.67) & 0.69 (0.38 to 1.20) & 1.01 (0.62 to 2.40) \\
Loose acetabular components (n = 4) & 0.08 (0.07 to 0.10) & 32.82 (23.32 to 41.29) & 0.76 (0.26 to 1.88) & 4.04 (1.75 to 13.65) \\
p-value & 0.861 & 0.940 & 0.900 & 0.027 \\
Tight screws (n = 23) & 0.08 (0.05 to 0.11) & 28.12 (13.40 to 45.61) & 0.57 (0.38 to 1.10) & 0.82 (0.51 to 1.11) \\
Loose screws (n = 12) & 0.09 (0.07 to 0.12) & 37.14 (23.33 to 48.57) & 0.89 (0.32 to 1.95) & 2.31 (1.75 to 4.65) \\
p-value & 0.263 & 0.161 & 0.797 & 0.002 \\
Inclination angle (°) & 0.09 (0.06 to 0.11) & 34.46 (17.67 to 49.82) & 0.73 (0.37 to 1.46) & 1.11 (0.74 to 2.49) \\
< 35° or > 45° (n = 8) & 0.08 (0.06 to 0.10) & 30.23 (16.51 to 39.92) & 0.51 (0.36 to 1.25) & 0.96 (0.57 to 4.23) \\
35° to 45° (n = 27) & 0.078 & 0.674 & 0.725 & 0.906 \\
\hline
\multicolumn{5}{l}{* IQR, interquartile range}
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\begin{table}[h]
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\caption{Correlation of details (mean, IQR\textsuperscript{*}) of the patients with volumetric articular wear (VAW) and backside volumetric change (BVC)}
\begin{tabular}{lcccc}
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& \textbf{Patient characteristics} & \textbf{Linear articular wear rate (mm/year)} & \textbf{VAW rate (mm\textsuperscript{3}/year)} & \textbf{BVC rate (mm\textsuperscript{3}/year)} & \textbf{BVC rate + screw-head indentations (mm\textsuperscript{3}/year)} \\
\hline
Age at time of implantation (yrs) & 62 (56 to 69) & -0.288 & -0.281 & -0.344 & -0.011 \\
Weight (kg) & 76 (71 to 84) & -0.077 & -0.101 & -0.136 & -0.247 \\
Height (cm) & 170 (165 to 176) & -0.100 & -0.079 & -0.245 & -0.368 \\
VAW rate (mm\textsuperscript{3}/year) & - & / & / & 0.632 & 0.724 \\
Time \textit{in vivo} (mths) & 54 (49 to 66) & 0.436 & 0.265 & 0.450 & \\
\hline
\multicolumn{5}{l}{* IQR, interquartile range}
\end{tabular}
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The rate of backside volumetric change was a median of 2.77% (IQR 1.36 to 3.97) of the rate of volumetric articular wear. The rate of volumetric articular wear (median 32.86 mm³/year, IQR 17.75 to 45.61), which was calculated from the linear articular wear rate (median 0.08 mm³/year, IQR 0.06 to 0.11), was significantly higher than the rate of backside volumetric change (median 0.69 mm³/year, IQR 0.38 to 1.20; sign test, p < 0.001, Table I).

In liners with screw-head indentations, the indentation volume accounted for a median of 1.10 mm³ (IQR 0.00 to 4.00) of the total backside volumetric change. The median rate of backside volumetric change plus the volume of the screw-head indentations was 1.10 mm³/year (IQR 0.63 to 2.76) and was significantly higher in patients with loose screws (Wilcoxon test, p = 0.002). Liners from acetabular components with loose screws had deeper screw-head indentations (median 0.45 mm, IQR 0.10 to 1.00) and a higher volume of indentations (median 8.35 mm³, IQR 0.0 to 24.0) than those from components with fixed screws (depth 0.20 mm, p = 0.002; volume 0.30 mm³, p = 0.004, Wilcoxon test Table I).

Increasing the size of the liner was associated with lower rates of backside volumetric change (52.0 vs 56.0 mm³/year, p = 0.023, 52.0 vs 60.0 mm³/year, p = 0.012, 56.0 vs 60.0, p = 0.013, Wilcoxon test), whereas the rates of volumetric articular change were not influenced significantly by the size of the liner (Table I). There was no clinically relevant or statistically significant difference in the rates of backside volumetric change and of volumetric articular wear in patients with an inclination angle of between 35° and 45° compared with those whose inclination angle was < 35° or > 45° (Table I).

There was no correlation between age, weight or height and the rates of volumetric articular wear or backside volumetric change. Correlation coefficients ranged from -0.344 to -0.079 (Table II). Gender had no influence on the rate of volumetric articular wear (p = 0.697) but male gender had a minor influence on the rate of backside volumetric change (Wilcoxon test, p = 0.054). Accordingly, the median rate of backside volumetric change in relation to the rate of volumetric articular wear was lower in men; (male (n = 30) 2.47% of the rate of volumetric articular wear; female (n = 5) 4.21%; Wilcoxon test, p = 0.069, Table I).

Manual workers (n = 16) had a tendency to a higher volumetric articular wear rate (median job-related burden vs no burden 35.06 vs 28.12 mm³/year, p = 0.589) and backside volumetric change rate (median job-related burden vs no burden 0.74 vs 0.51 mm³/year, Wilcoxon test, p = 0.909).

Patients with migration of the acetabular component > 1 mm (n = 12) had a higher backside volumetric wear rate (Wilcoxon test, p = 0.073) and volumetric articular wear rate (p = 0.263) than those with migration of < 1 mm (n = 23). The four loose acetabular components had a median migration of 4.25 mm (IQR 1.9 to 9.6) whereas the stable components (n = 31) had a median migration of 0.9 mm (IQR 0.7 to 1.3 mm, Wilcoxon test, p = 0.01). The volumetric articular wear rate in patients with stable components (median 30.23 mm³/year) was lower than that in the four loose acetabular components (median 32.82 mm³/year; Wilcoxon test, p = 0.94). The median backside volumetric change rate plus the volume of the screw-head indentations was significantly lower in patients with stable acetabular components (median 1.01 mm³/year) compared with those with loose components (median 4.04 mm³/year, Wilcoxon test, p = 0.027). Loose acetabular components had significantly deeper screw-head indentations than stable components (median 0.65 mm vs 0.20 mm, Wilcoxon test, p = 0.005).

**Discussion**

Articular wear rates for different articulation pairings have been investigated, but to our knowledge volumetric backside change has only been quantified by Kurtz et al. Our study introduced a new method of measuring the rate of backside volumetric change.
We note its limitations, including unequal gender distribution and a low number of loose acetabular components which could have influenced some of the results. Measurements were all performed on liners of the same design and the conclusions may not be applicable to those of different designs. However, as there are no published direct measurements of the backside volumetric change our values give an estimate of the magnitude of true backside change.

Our measurements of volumetric articular wear ranged within what is considered ‘normal’ for polyethylene-ceramic pairings. Comparison studies on wear rates of ceramic and metal femoral heads have shown large variations. Considering that we used a direct-wear measurement instead of a radiological method and that others have applied different formulae to calculate volumetric articular wear, our wear rates for the head-liner interface correlated instead of a radiological method and that others have applied different formulae to calculate volumetric articular wear, our wear rates for the head-liner interface correlated with published data for 28 mm and 32 mm wear, whereas Kurtz et al showed volumetric backside wear rates of the median rate of volumetric articular wear, whereas Kurtz et al and 37 showed volumetric backside wear rates of between 0.11% and 0.17% of the volumetric articular wear, depending on the conformity of the shell and number of screw-holes. However, their study used a different design, polished interface conditions and initial wear rates in their analysis. Both studies showed that articular wear predominated over backside change. However, our direct measurements showed that backside volumetric change was at least ten times larger than has been suggested. Moreover, their computer model did not take creep into account. Our values included creep and wear which might be responsible for the difference, but our method of analysis did not allow us to distinguish between them.

Several reports have described the inverse relationship between liner size and articular wear. Likewise, but not statistically significant, we found a decrease in the volumetric articular wear with increasing size of liner. However, the relationship was statistically significant for backside volumetric change. A retrieval study by Yamaguchi et al described similar findings, and suggested that liners with backside deformation were significantly thinner than those without, supporting the experimental finding that creep is lower for thicker polyethylene. Increasing size and thickness of the liner results in greater stress distribution and less plastic deformation.

Articular wear correlated significantly with the backside volumetric change. This is not surprising, since a higher volumetric articular wear increases the general load on the liner, thereby leading to higher creep.

As in previous studies, we found no correlation between the volumetric articular wear and weight or age, possibly because our study group was too homogenous, with only small deviations from the mean. Livermore et al found that the volumetric articular wear increased with increasing weight, whereas Devane et al reported increased volumetric articular wear with greater activity, but no correlation with body-weight. We also observed a tendency to increased volumetric articular wear and backside volumetric change in manual workers.

Migrated acetabular components showed a significantly higher rate of backside volumetric change and screw-head indentations than non-migrated components (p = 0.011). While migration and wear analysis of the articulation side were calculated by radiological methods backside volumetric change cannot be measured by these methods because of its small amount compared with the volumetric articular wear (2.8%). We have not found any reports on migration with regard to backside volumetric change but it is not surprising that migrated acetabular components with screws will produce more backside volumetric change than non-migrated components. In our series, loose components with larger migration (median 4.25 mm) had a significantly higher backside volumetric change along with screw-head indentations than stable components (p = 0.027).

Yamaguchi et al excluded liners with screw-head indentations whereas, in our study, such liners were also evaluated to investigate the role of screws in backside wear. In order to show their influence clearly, the volume of the screw-head indentations was not incorporated in the volume of the backside volumetric change. The results suggested that the volume of the screw-head indentations might be higher than the backside volumetric change if the depth of the indentations was high. This was observed significantly more often in cases in which loose screws were detected during revision surgery. The median rate of backside volumetric change plus the volume of the screw-head indentations was significantly lower in patients with stable acetabular components compared with those with loose components. The higher levels in the latter might also have been due to a longer time in vivo (median loose acetabula 73.5 months vs fixed acetabula 53.0 months).

Ten of the 12 liners with loose screws showed screw-head indentations with partially intact or obliterated machining marks, indicating evidence of initial wear. Two liners with loose screws had no indentations. In these cases there must still have been a space between the screw-head and the backside surface of the liner, despite loosening of the screw. Several studies have shown the importance of screws in connection with backside changes. Impingement of screw-heads against the liner can lead to backside wear but countersinking the screw-holes in the shell should be sufficient to prevent this. In the case of a loose acetabular component, movements between it and the liner will lead to wear because of friction against the stable screw-head.

As in earlier reports we found a positive correlation between the volumetric articular wear, backside volumetric change and time in vivo, but no association between weight...
or age and the backside volumetric change. Yamaguchi et al.13 found a relationship between male gender and screw-hole deformation which could also be interpreted as a correlation between greater body-weight and backside deformation. With only five women, the gender distribution in our group was not homogenous and could limit our results on gender-related polyethylene wear.

The stability of the screws did not affect either the volumetric articular wear nor the rate of backside volumetric change. No reason was found to indicate why the state of the screws should influence articular wear, but our study considered the interface screw-head/liner to be an additional interface at the backside. Under certain circumstances this interface might have had greater influence on the generation of polyethylene wear particles on the backside than the liner-shell interface itself. The locking mechanism is likely to be the most important factor in backside change at the liner-shell interface. Several in vivo models have shown that a rigid locking mechanism limits micromovement and therefore backside wear,19-21,45 whereas polishing the inner side of the shell does not seem to reduce such wear significantly.12,21 Since the rate of backside volumetric change in our acetabular components was reasonably small, the locking mechanism was considered to be adequate.

The method for determining backside volumetric change is applicable to any design. It estimates the true amount of the backside volumetric change and is suitable for comparison with values from the articulation side.

Provided that the acetabular component is stable and that the screws are well inserted and anchored, backside wear should not be a problem. Impingement of screw-heads against the liner could potentially produce more wear particles than those generated by the contact between the liner and the shell. Because the rate of backside volumetric change is only 2.8% of the rate of volumetric articular wear and with creep likely to be a significant contributor to the rate of backside volumetric change, the amount of debris generated by wear at the backside of the liner may not be sufficient to create a strong osteolytic response.

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References


