INFLUENCE OF PROSTHESIS MIGRATION IN HUMAN HIP JOINT USING MULTI-BODY SIMULATION

Stefanie Betancur Escobar, betancur@ifum.uni-hannover.de
Leibniz Universitaet Hannover, Institute for Metal Forming and Metal Forming Machine, Department of CA-Technologies, Hannover, Germany
Anas Bouguecha, bouguecha@ifum.uni-hannover.de
Leibniz Universitaet Hannover, Institute for Metal Forming and Metal Forming Machine, Department of Numerical Methods, Hannover, Germany
Ingo Nolte, ingo.nolte@tiho-hannover.de
University of Veterinary Medicine Foundation Hannover, Small Animal Clinic, Hannover, Germany
Christina Stukenborg-Colsman, villoe.pal@ddh-gruppe.de
Hannover Medical School, Department of Orthopaedics, Hannover, Germany
Bernd-Arno Behrens, behrens@ifum.uni-hannover.de
Leibniz Universitaet Hannover, Institute for Metal Forming and Metal Forming Machine, Hannover, Germany

Abstract. *The implantation of a total hip prosthesis is a routine procedure, which is performed because of advanced hip joint damage in human medicine as well as veterinary medicine in dogs. The long-term result of a hip prosthesis is mainly determined by migration or aseptic loosening of the implant. The migration and loosening of the prosthesis respectively is still a current problem and may be caused by stress shielding in the periprosthetic bone and abrasion particles of the tribological pairing head/cup. Due to pain and late complications (e.g. restricted movement), an advanced loosening leads to a cost-intensive revision operation with the change of the endoprosthesis. To improve the design of the prosthesis and thus to avoid a revision operation, it is essential to know the forces acting on the hip joint. Therefore, Multi-Body Simulations (MBS) and Finite Element Method (FEM) are used. MBS are necessary for estimating the forces acting on the hip joint with different body movements. Afterwards these computed hip forces can be used for FE analyses in order to determine the strain adaptive bone remodelling due to different loading situations. The overall aim of this project is the development and establishment of a simulation based method for calculating these effects. In this context, a MBS model of a human woman was generated. By means of this model, the hip joint loadings during the human gait cycle have been calculated. In order to perform the simulation, the kinematic data of a clinically healthy female test subject were determined by gait analysis. These data were implemented into the calculation to drive the static MBS model. The anthropometric parameter of the generated model was based on the female test subject. Data resulting from ground reaction forces during walking were obtained from force plate measurements and were imported in the MBS model. In addition to the loads of the periprosthetic supplied hip joint, the loads of a migrated prosthesis were determined. This analysis showed that a low migration of the prosthesis had an impact on the load collective in the periprosthetic supplied hip joint. The influence of the varied load collective must be considered in the FE model which is already developed for the determination of the strain-adaptive bone remodelling. This change of load leads to a new contact condition and thus to a modified bone remodelling, so that a coupling of FEM and MBS is unavoidable.*

Keywords: *Multi-Body Simulation, migration, hip joint forces, hip prosthesis*

1. INTRODUCTION

The treatment of severe hip joint damage with total hip prosthesis (THP) is established as routine surgery in human medicine as well as veterinary medicine in dogs. Despite long-term experience with endoprosthetic supply of hip joints, the migration and loosening of the prosthesis respectively is still a current problem and may be caused by stress shielding (Kuiper and Huiskes, 1997) in the periprosthetic bone or abrasion particles of the tribological pairing head/cup (Ingham and Fisher, 2000). In addition to the loosening of the prosthesis stem, the loosening of acetabular components (cup) has a significant influence on the life of the THP (Garcia-Cimbrelo *et al.*, 2000), (Wright *et al.*, 2001), (Grant and Nordsletten, 2004), (Shetty *et al.*, 2006), (Laursen *et al.*, 2007). Due to pain and late complications (e.g. restricted movement), an advanced loosening leads to a cost-intensive revision operation with the change of the endoprosthesis. To improve the design of the prosthesis and thus to avoid a revision operation, it is essential to know the forces acting on the hip joint. Therefore, Multi-Body Simulations (MBS) and Finite Element Method (FEM) are used. MBS are necessary for estimating the forces acting on the hip joint with different body movements. Afterwards these computed hip forces can be used for FE analyses in order to determine the strain adaptive bone remodelling due to different loading situations.

The overall aim of this project is the development and establishment of a simulation based method for calculating these effects. Within this project, a FE model has already been developed to predict the change in load distribution and hence to estimate the resulting bone remodeling in the periprosthetic femur and acetabulum respectively (Behrens *et al.*, 2008a) (Behrens *et al.*, 2008b), (Behrens *et al.*, 2009), (Bouguecha *et al.*, 2010). For this analysis, the boundary conditions (forces acting on the hip joint) were obtained from the clinical investigations of Bergmann et al. (Bergmann *et al.*, 2001). In the current study, a human MBS model was generated to determine the hip joint loadings for a migrated prosthesis during the human gait cycle. Apart from the determination of influence of a migration or loosed prosthesis on the bone remodelling a coupling of FEM and MBS is the long term goal.

2. MATERIALS AND METHOD

The MBS model was generated using the commercial software BGR.LifeMOD 2010.0.0™ (Biomechanics Research Group, Inc., USA) which is based on the software MSC.ADAMS® (Mechanical Dynamics Inc., USA). Previously, gait analysis on an instrumented treadmill was performed to determine the kinematic and kinetic data for the modelling.

2.1 Determination of the kinematic and kinetic data

The gait analysis was realized at the gait laboratory of the clinic for small animals at the University of Veterinary Medicine Foundation Hannover. This laboratory is equipped with 4-infrared-camera system (Vicon MX-3+, Oxford Metrics, USA) and an instrumented treadmill with four integrated 3D-force plates (Bertec CTM4-B07, Columbus-OH, USA) placed under four particular straps. The four straps of the treadmill are driven synchronously. Consequently it is possible to examine human or canine probands.

For the measurement of the required kinematic data, a clinically healthy female test subject (BW: 65 kg, height: 1.63 m, age: 24 years) was labeled with 16 retro reflective markers (diameter: 16 mm, 8 per side) with double-side adhesive tape. The marker positions based on the Plug-in-Gait Marker set for the Lower Body (LifeMOD, 2011) are shown in Fig.1 and are described more precisely in Tab.1.

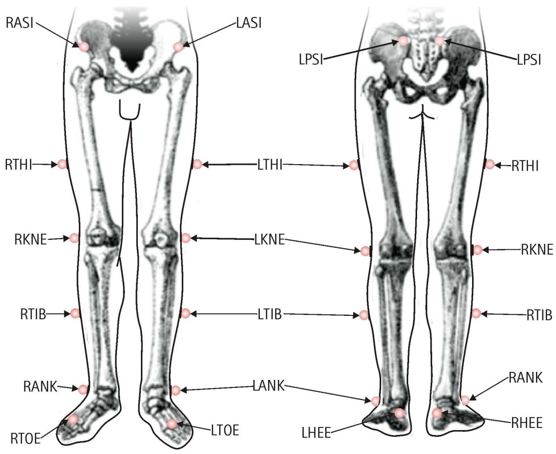

Figure 1. Plug-in-Gait Marker Placement Protocol (LifeMOD, 2011)

Table 1. Precise position of the Plug-in-Gait Marker

Segment	Marker	Position
pelvis	LASI/RASI LPSI/RPSI	anterior superior iliac spine posterior superior iliac spine
femural	LTHI/RTHI LKNE/RKNE	over the lower lateral 1/3 surface of the thigh lateral epicondyle
tibial	LTIB/RTIB LANK/RANK	over the lower 1/3 surface of the shank lateral malleolus
foot	LTOE/RTOE LHEE/RHEE	second metatarsal head calcaneous

After calibration of the system (mean measurement error for all four cameras: 0.07 mm), motion capture of the female test subject during gait on two straps of the treadmill was executed. The strap speed was 1.1 m/s. A trail of 10 sec duration was recorded to achieve enough repeatable motion data. Sampling rate for the kinematic data was 100 Hz, whilst the kinetic data from the treadmill were recorded with a frequency of 1000 Hz. By means of the software Vicon Nexus®(Oxford Metrics, USA), the measured trail data were recorded, processed and then exported to an MS-excel compatible file type (ASCII). This trail file included coordinate information in x-, y- and z-direction for each of the markers for the whole measurement duration. Furthermore the ground reaction forces and moments respectively as wall as the force application points in the three coordinate directions were written in the trail file.

2.2 Creation of the MBS model

The bony structures of the MBS model are created automatically from an anthropometric database generating segment dimensions, mass and inertia tensor. The database used is called GeBOD and creates a human model based on simple description (gender, age, height and weight) which can be modified. For the conducted simulation, only the lower torso was generated. The anthropometric parameter of the generated model was based on the female test subject. They were implemented into the calculation to drive the static MBS model. After the creation of the bony structures, the joints were modeled. The hip and ankle were modeled as joints with three rotational degrees of freedom (DOF). The knee, however, was modeled as a joint with one rotational DOF in the medio-lateral axis. The implementation of the muscle set was generated automatically by BGR.LifeMod. The default muscle set for each leg has 45 muscles, which are robust for modelling gait analysis (LifeMOD, 2011).

The last step of the creation of the MBS model was adding the artificial hip joint. Therefore the right hip joint was deleted and replaced by the prosthesis, which was imported as a shell file. The fixation of the prosthesis in the related bone was realized using bushing element forces, whereas the physiological rotation center of the hip joint was reconstructed to determine the right hip forces. A bushing force was also used to simulate the interaction between the stem ball and the cup. In Figure 2 the created MBS model with the musculo-skeletal structures as well as the joints and the “implanted” prosthesis is shown.

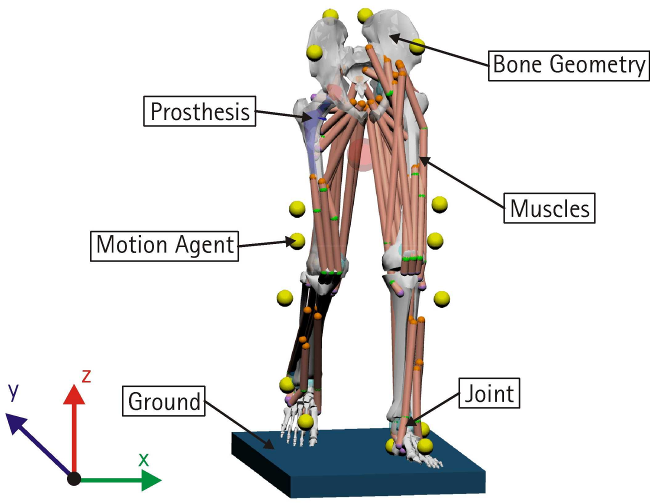

Figure 2. Created MBS Model of the feminine test subject.

2.3 Integration of kinematic and kinetic data

Before the integration of the measured kinematic and kinetic data, the written trail file was converted in a “Standard Linear Format” file (SLF). This file type can contain information on units, anthropometrics, joints and posture or motion capture data. In this case two SFL files were generated containing the kinematic and kinetic data respectively by using a self written VBA-Macro. These SLF files were imported into the previously generated model.

By means of the kinetic SLF files, the ground reaction forces, moments and the force application points were integrated in the model. On the basis of the kinematic SLF file Motion Agents (Fig. 2) were created. These Motion Agents were placed at the same position as the used Plug-In gait marker set. The function of the Motion Agents is to position the model in the initial Posture and after that to move the model using the measured trajectories of the retro reflective marker.

2.4 Simulation

Based on the known measured motion and force data, a so-called inverse dynamic simulation was performed. During this simulation the motion agents move according to the measured motion capture data, which causes a movement of the MBS model. Thus, each muscle is trained in order to accomplish the required shortening/lengthening pattern to execute the intended movement. The definition of trained means in this context, that the contraction histories are recorded.

After the inverse, the forward dynamic simulation was performed. For this simulation, the recorded muscle contractions were used to create torque and forces so as to produce the proper internal reaction needed to move the model with the motion from the inverse dynamic simulation. A PID-controller (Proportional-Integral-Derivative) minimizes the error between the recorded contraction histories of the muscles and the actual muscle values received from the forward dynamic simulation.

Subsequently the simulation with the "healthy" model, a total hip endoprosthesis was implemented and the inverse and forward simulation respectively was repeated. The position of the prosthesis was selected such that the rotation center was reconstructed. By means of this model, the prosthesis migration proximally as well as proximally with an anditional anterior and posterior part was examined. For the changed position, the maxima of the normalized hip resultant of the whole gait cycle were compared with the values of the prosthesis placed in the rotation center.

3. RESULTS

For the validation of the MBS model, a comparison of the simulation with the measurement results of the workgroup of Bergmann (Bergmann *et al.*, 2001) (blue cuves) and the free public database Orthoload (Orthoload, 2011) (green curves) was executed. This comparison shown in Fig. 3 indicate that the calculated results with the MBS model have a similar pattern and magnitude of force compared with those recoded by Bergmann et al. and Orthoload.

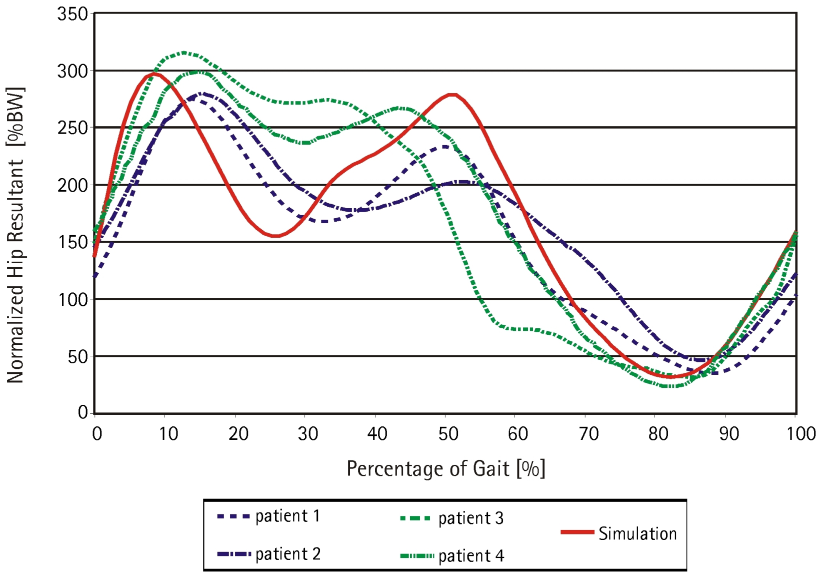

Figure 3. Comparison of simulated and measured Hip Resultant.

As mentioned above, the prosthesis migration proximal as well as proximal with an anterior and posterior part was examined. For all directions the value of the displacement is 2 mm. This comparison is shown in Fig. 4 which contains force progression of the normalized (in terms of the body weight (BW)) hip resultant of the prosthesis positioned in the rotational center and of the same migrated prosthesis in the three different directions.

Compared with the prosthesis in the rational center, a proximal migration of the prosthesis leads to reduction of the first maximum by 3 % and by 2 % of the second maximum. An additional posterior part causes an increase of the first maximum by 3 % as well as reduction of the second maximum by 6 %. However, an additional anterior part leads to a reduction of the first maximum by 5 % and an increase of the second maximum by 8 %.

Apart from the results already presented, the effects of a medial, cranial, posterior and anterior migration are specified in Tab. 2, so that the influence of each single direction can be estimated. In this case, the value of the migration is also 2 mm. The influence of a migration only in cranial and medial direction respectively is much less than for the combination of both. Whereas cranial migration leads to a reduction of both maxima by 0.2 %, a medial migration causes a reduction of the first maximum by 1 % and a reduction of the second maximum by 0.2 %.

The increase or decrease of the maxima after a migration of the prosthesis only in anterior or posterior direction is generally higher compared to the results from the combination with an additional proximal part. This excluded the reduction of the first maximum after a migration anterior. With a decrease by 4.5 %, the change in the first maximum is lower compared to the result of the combination.

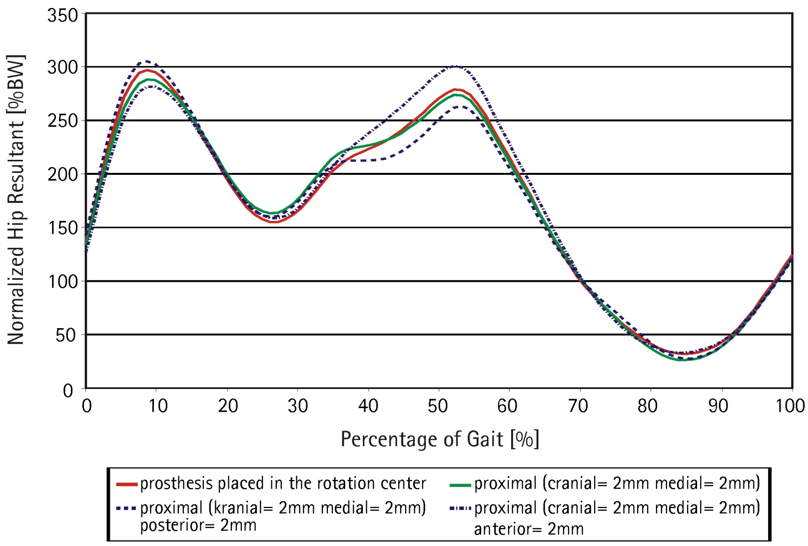

Figure 4. Influence of the prosthesis migration on the force progression of the normalized hip resultant

Table 2. Influence of the prosthesis migration on the maxima compared to the values of the prosthesis placed in the rotation center.

Anatomical direction	**Max 1**	**Max 2**
proximal (cranial + medial)	-3%	-2%
proximal + posterior	3%	-6%
proximal + anterior	-5%	8%
cranial	-0.2%	-0.2%
medial	-1%	-0.2%
posterior	5%	-6.5%
anterior	-4.5%	10%

Value of the migration in each direction $= 2\ mm$

4. DISCUSSION

In the study presented here, a MBS model based on the anthropometric data of a clinically healthy female test subject were created to determine the loads of a periprosthetic supplied hip joint and migrated prosthesis respectively. Therefore a gait analysis was performed, by means of which the kinematic and kinetic data of the test subject were determined.

In previous studies simulations of the musculoskeletal loading conditions in the hip joint were already executed (Heller *et al.*, 2001), (Manders *et al.*, 2008). Heller et al. (Heller *et al.*, 2001) determined the musculoskeletal loading conditions during walking and stair climbing for a number of patients based on the telemetric hip force measurements and individual lower extremity models. In all cases, there was a good agreement between the in vivo measured and calculated hip contact forces. In contrast to this direct validation, a direct validation with in vivo data was not possible in the present study because those data of the test subject were not available. On this account, the validation of the MBS model was conducted with the comparison of the simulation and the measurement results of Bergmann et al. (Bergmann *et al.*, 2001) and the free public database Orthoload (Orthoload, 2011) so that a qualitative comparison was performed. This qualitative comparison shows a good correlation between the simulation and the measurement.

The investigation of the migration shows that a low migration of the prosthesis may have a considerable influence on the load collectives. It could be demonstrated, that a proximal migration causes a higher increase and decrease respectively compared with the results for the migration only in cranial or medial direction. Furthermore, it could be determined that an additional migration direction posterior and anterior respectively strongly influenced the load collective in the periprothetic hip. However, the increase or decrease of the maxima is generally lower than for a migration only in anterior-posterior direction, so that the proximal part leads to a reduction of the influence on the load collective.

Besides different investigations with the developed simulation model, Heller et al. already carried out a study focusing on the influence of the cup position on the loads in the hip joint (Heller *et al.*, 2007). As shown in the present study, Heller et al. also determined that the position of the cup and rotation center respectively has an influence on the load collective in the periprosthetic hip. Thereby the mean and maximum joint forces were determined. Certainly a direct comparison of the values is not possible, due to the fact that Heller et al. only specified the results of the mean forces. The results of the maximum forces should have comparable results but were not refined in the investigation. In summary, it can be stated, however, that the results achieved in this study shows the same tendency compared with the values of the Heller study.

The influence of the varied load collective determined in this study must be considered in the FE model which is already developed for the determination of the strain adaptive-bone remodelling. This change of load leads to a new contact condition and thus to a modified bone remodelling. Consequently, in future studies, a coupling of the MBS model and the FE model have to be realized.

5. ACKNOWLEDGEMENTS

The study was realized in the framework of the collaborative research center (SFB) 599 "Sustainable degradable and permanent implants made of metallic and ceramic materials". The authors would like to thank the German Research Foundation (Deutsche Forschungsgemeinschaft, DFG) for its financial support of the project TP 06 "Numerical simulation on stress-compatible design of total prosthetic joint replacements and implants".

6. REFERENCES

Behrens, B.A., Nolte, I., Wefstaedt, P., Stukenborg-Colsman, C. and Bouguecha, A., 2008a. "Finite element analysis of bone remodeling after hip resurfacing arthroplasty". *ECIFMBE 2008*, Vol. 4.

Behrens, B.A., Nolte, I., Wefstaedt, P., Stukenborg-Colsman, C. and Bouguecha, A., 2009. "Numerical investigations on the strain-adaptive bone remodelling in the periprosthetic femur: Influence of the boundary conditions". *BioMedical Engineering OnLine*, Vol. 8, pp. 1–9.

Behrens, B.A., Wirth, C.J., Windhagen, H., Nolte, I., Meyer-Lindenberg, A. and Bouguecha, A., 2008b. "Numerical investigations of stress shielding in total hip prostheses". *Journal of Engineering in Medicine*, Vol. 222, pp. 593–600.

Bergmann, G., Deuretzbacher, G., Heller, M., Graichen, F., Rohlmann, A., Strauss, J. and Duda, G.N., 2001. "Hip contact forces and gait patterns from routine activities". *Journal of biomechanics*, Vol. 34, No. 7, pp. 859–871.

Bouguecha, A., Elgaly, I., Stukenborg-Colsman, C., Lerch, M., Nolte, I., Wefstaedt, P., Matthias, T. and Behrens, B.A., 2010. "Numerical investigations of the strain-adaptive bone remodeling in the prosthetic pelvis". *IFMBE Proceedings*, Vol. 29.

Garcia-Cimbrelo, E., Diaz-Martin, A., Madero, R. and Munera, L., 2000. "Loosening of the cup after low-friction arthroplasty in patients with acetabular protrusion: The importance of the position of the cup". *The Journal of bone and joint surgery. British volume*, Vol. 82, No. 1, pp. 108–115.

Grant, P. and Nordsletten, L., 2004. "Total hip arthroplasty with the lord prosthesis: A long-term follow-up study". *The Journal of bone and joint surgery. American volume*, Vol. 86-A, No. 12, pp. 2636–2641.

Heller, M.O., Bergmann, G., Deuretzbacher, G., Dürselen, L., Pohl, M., Claes, L., Haas, N.P. and Duda, G.N., 2001. "Musculo-skeletal loading conditions at the hip during walking and stair climbing". *Journal of biomechanics*, Vol. 34, No. 7, pp. 883–893.

Heller, M.O., Schröder, J.H., Matziolis, G., Sharenkov, G., Taylor, W.R., C., P. and G.N., D., 2007. "Musculoskeletal load analysis: A biomechanical explanation for clinical results – and more?" *Der Orthopäde*, Vol. 36, No. 3, pp. 188–194.

Ingham, E. and Fisher, J., 2000. "Biological reactions to wear debris in total joint replacement". *Proceedings of the Institution of Mechanical Engineers. Part H, Journal of engineering in medicine*, Vol. 214, No. 1, pp. 21–37.

Kuiper, J.H. and Huiskes, R., 1997. "The predictive value of stress shielding for quantification of adaptive bone resorption around hip replacements". *Journal of biomechanical engineering*, Vol. 119, No. 3, pp. 228–231.

Laursen, M.B., Nielsen, P.T. and Søballe, K., 2007. "Bone remodelling around ha-coated acetabular cups: A dexa study with a 3-year follow-up in a randomised trial". *International orthopaedics*, Vol. 31, No. 2, pp. 199–204.

LifeMOD, 2011. LifeMOD Tutorial. 01.03.2011 <http://www.lifemodeler.com/LM_Manual_2010/>.

Manders, C., New, A. and Rasmussen, J., 2008. "Validation of musculoskeletal gait simulation for use in inverstigation of total hip replacement". *Journal of Biomechanics, Supplement 1*, , No. 41, p. 499.

Orthoload, 2011. Orthoload Database, 01.03.2011, <www.orthoload.com>.

Shetty, N.R., Hamer, A.J., Kerry, R.M., Stockley, I., Eastell, R. and Wilkinson, J.M., 2006. "Bone remodelling around a cemented polyethylene cup: A longitudinal densitometry study". *The Journal of bone and joint surgery. British volume*, Vol. 88, No. 4, pp. 455–459.

Wright, J.M., Pellicci, P.M., Salvati, E.A., Ghelman, B., Roberts, M.M. and Koh, J.L., 2001. "Bone density adjacent to press-fit acetabular components: A prospective analysis with quantitative computed tomography". *The Journal of bone and joint surgery. American volume*, Vol. 83-A, No. 4, pp. 529–536.

7. Responsibility notice

The authors are the only responsible for the printed material included in this paper