AMAS Workshop on Orthopaedic Biomechanics
OBM'02 – (pp.33–45) – Wrocław, September 1-4, 2002.

Biomechanics of the hip joint

G. BERGMANN, F. GRAICHEN, and A. ROHLMANN

Biomechanics Laboratory
Charité – University Medicine Berlin
Campus Benjamin Franklin
Hindenburgdamm 30, 12203 Berlin, Germany
bergmann@biomechanik.de
www.biomechanik.de

Key words: *Hip joint, force, telemetry, biomechanics*

1. Introduction

Biomechanics of the hip joint involves knowledge of forces in the joint and surrounding musculature, the design of hip joint replacement and many other aspects. Here the focus is set on the contact forces acting at the joint during various everyday activities. Such data are important for improving hip endoprostheses and their fixation in bone, for advising patients about which activities may be disadvantageous for their artificial joint and for obtaining information about the best operative procedures.

It has long been known that the forces in the hip joint are higher than the body weight during walking and most activities performed while standing on one leg (Pauwels 1935, 1973). The high forces are caused by the weight of the upper body acting at a long lever arm relative to the supporting hip joint. The muscles required to maintain the balance of moments always have much shorter lever arms. Their force must therefore exceed the upper body weight. The muscles then transmit their force through the joint across which they act. Simple models of walking and standing are insufficient for describing complex dynamic loading conditions. Sophisticated analytical models have been designed to simulate the three-dimensional activities of all muscles acting across the hip joint. They include gait analysis with computer simulations of the dynamic equilibrium of forces and moments. However, these models and methods are so complex that the calculated hip contact forces (Crowninshield et al. 1978, Brand et al. 1994) long remained uncertain. It has just recently become possible to improve the analytical methods by directly comparing exactly measured contact forces with those calculated (Heller et al. 2001, Stansfield et al. 2003).

To obtain realistic data for contact forces in the hip joint, hip implants were instrumented with built-in load sensors and telemetric data transmission (Bergmann et al. 1988, Graichen et al. 1988, 1999). Such force-sensing implants were inserted in 9 joints of 7 patients, and measurements were taken during most common activities of everyday life for up to 9 years (Bergmann et al. 1995 to 2002). This paper summarizes the most interesting findings.

Stable hip implant fixation does not seem to always require the lowest possible contact forces. Forty years of clinical experience have shown that well-designed hip endoprostheses correctly implanted either with or without bone cement can transfer 'normal' loads even for decades. On the other hand, one can never be absolutely certain whether loosening of a hip implant has already started and may gradually increase. We therefore advise implant recipients against activities that frequently load the joint with 'excessive' forces. We regard contact forces higher than during fast walking as 'excessive' but are aware that such suggestions are somewhat arbitrary. They are certainly on the 'safe side', however. Laboratory experiments have demonstrated that especially backwards rotation of hip implants may endanger their fixation stability (Tanner et al. 1988, Nunn et al. 1989, Burke et al. 1988). Thus one goal of telemetric measurements was to determine whether the in vivo rotation moments may be high enough to cause implant loosening (Bergmann et al. 1995).

2. Methods

Two kinds of clinically established total hip implants were modified (Fig. 1) to incorporate electronics which measured the 3 components of the contact force acting between the head and ball of the implants (Graichen et al. 1988, 1999). All implants had ceramic heads and polyethylene cups, except one with a ceramic cup. The first model was implanted with bone cement; the second was fixed without cement. The latter additionally measured the temperature distribution in the implant to determine whether friction-induced heating of the implant could endanger its fixation in bone. A total of 9 instrumented hip joints were implanted in 7 patients.

Data monitored in real time on a PC were additionally videotaped with synchronous recording of patients' images. This allowed detailed analyses of the data later on. The contact forces were measured relative to the femur and are expressed in percent of the patient's body weight (% BW). Since they

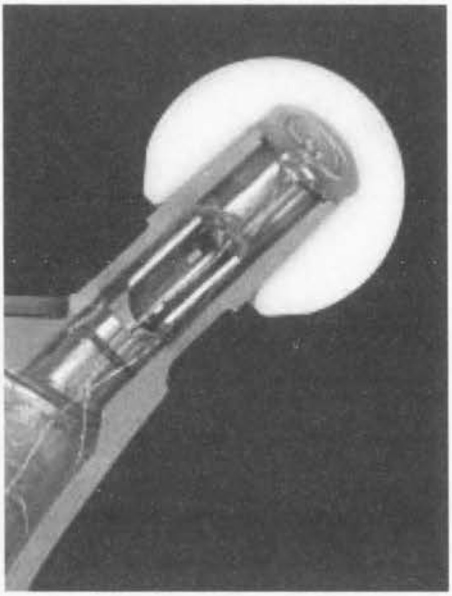

FIGURE 1. Instrumented hip implant, model 2.
Three load sensing strain gauges and 9 temperature sensors are incorporated in the implant with inductive power supply. Two telemetries transmit the signals. Measured are the three components of the hip contact force and the temperature distribution along the implant.

vary individually, we first averaged the force patterns vs. time from several trials of the same patient. Such data from single and average trials is used in the diagrams. The numbers cited in the text are ranges or averages of peak values from several (averaged) subjects. The numbers denote resultant forces that comprise the 3 spatial components. Using anatomic data from CT's, the moments acting around the stem of the implant could also be calculated but are only mentioned here if they are of special interest.

Four of the patients had synchronous recording of their gait data, EMG and telemetric contact forces during standard activities like standing, walking, going upstairs or standing up (Bergmann et al. 2001c,d). Using the contact forces as a 'gold standard', this database was taken to improve the analytical methods applied to calculate the muscle forces acting across the hip joint (Heller et al. 2001, Stansfield et al. 2003).

3. Results

When ***standing*** on two legs, the contact force had an average magnitude of 70% BW and up to 99% BW were found in one of the patients (Fig. 2).

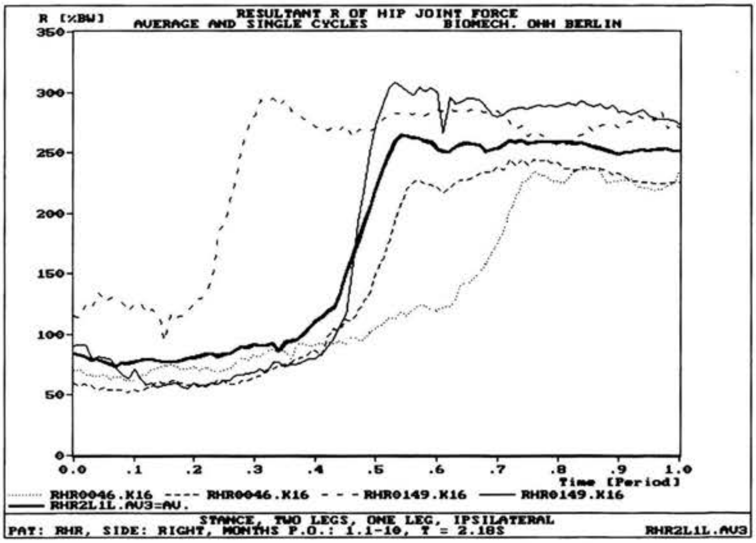

FIGURE 2. Standing.
Resultant contact forces from 4 trials of 1 patient (thin) and their average (thick). Scale in percent of body weight (% BW). The patient first stands on 2 legs and then shifts the weight to one leg. Average data from several subjects were averaged again to find contact forces of a 'typical' subject.

As mentioned, the forces from single trials, shown in the diagrams, may vary from the averaged data. The joint force tends to decrease slightly with increasing foot distance.

When standing on one leg (Fig. 2), average forces of 25% BW and individual maxima of 327% BW are measured. This is in the range Pauwels (1973) found using a very basic static model.

Maximum forces for slow ***walking*** are nearly the same or only slightly higher than for standing on one leg (Fig. 3). At a speed of 3 km/h the average peak force was 270% BW and the individual maximum was 321% BW. This was also predicted by Pauwels. The force is always low during the swing phase. It increases before the foot touches ground and reaches a value of about 100 to 150% BW on impact. Force peaks at the instant of heel strike, like those transmitted from the ground to the foot, were never observed at the hip, even at higher speed and when hitting the ground hard.

Jogging at 7 km/h was the fastest exercise investigated in one patient and led to peak loads of 550% BW. These were the highest values measured for any of the investigated activities except stumbling.

Higher loads would be expected for going ***upstairs*** because of the increased hip joint flexion and the difficulty elderly subjects often have when

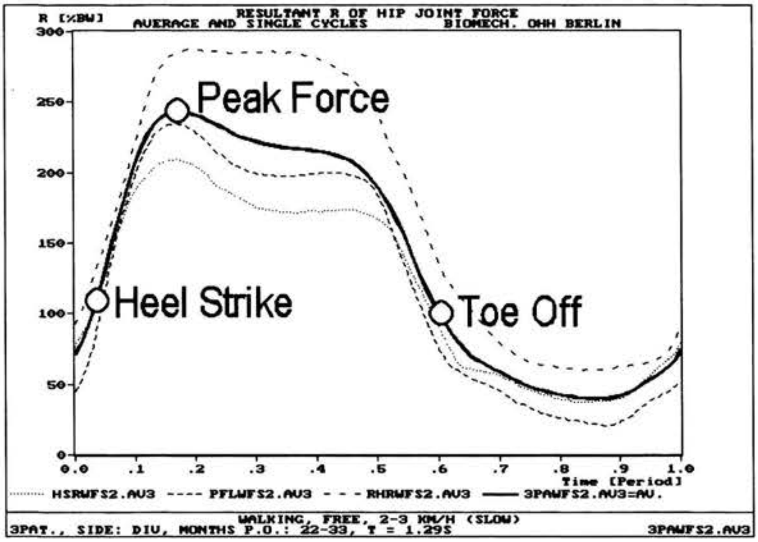

FIGURE 3. Slow walking.
Resultant contact forces from 3 patients (thin) and average (thick). Shown are single steps at 2 to 3 km/h. The average peak force is about 240% BW. During the swing phase the load is low. The instant of heel strike (marked) precedes the instant of peak force.

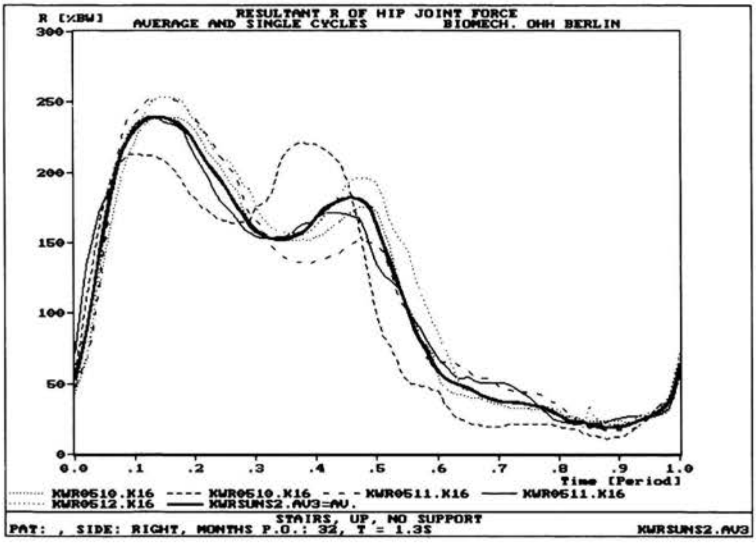

FIGURE 4. Walking upstairs.
Resultant forces from 5 steps of 1 patient (thin) and their average (thick).

using a staircase (Bergmann et al. 1995). However, the contact force for going upstairs is only slightly higher than for walking on level ground (Fig. 4): average loads are at 283% BW with individual values up to 370% BW. The force component –Fy acts backwards at the implant head and thus determines most of the torsional moment aimed at turning the implant around the stem axis. This component and the torque are higher for climbing stairs than for walking on level ground but remain within the range found for fast walking.

Sitting down and standing up cause lower contact forces than walking because the upper body is supported by both legs (Bergmann et al. 2001c,d). Peak forces had average values of 192% BW with individual maxima of 281% BW when standing up (Fig. 5). The hip contact force is lowered by about 30% in the case of hand-to-arm-rest support of the upper body. Hand support at the knees can reduce the joint force only slightly.

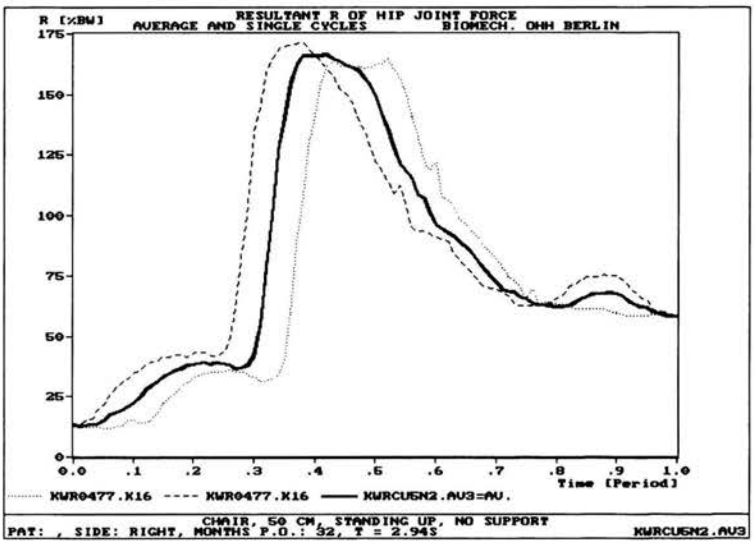

FIGURE 5. Standing up.
Resultant forces from 2 trials of 1 patient (thin) and their average (thick). The subject first sit, then stand up and finally stands on both feet.

Even activities in lying positions without external loads can cause high forces in the hip joint. As during sitting, the hip joint is nearly unloaded when ***lying relaxed***. However, when ***lifting the straight leg*** by 45° in a supine position, the contact forces rises to 139% BW on average and even reaches 186% BW (Fig. 6). The force at the contralateral side will increase to the 129% BW on average, i.e. nearly the same value (Fig. 7). Obviously both

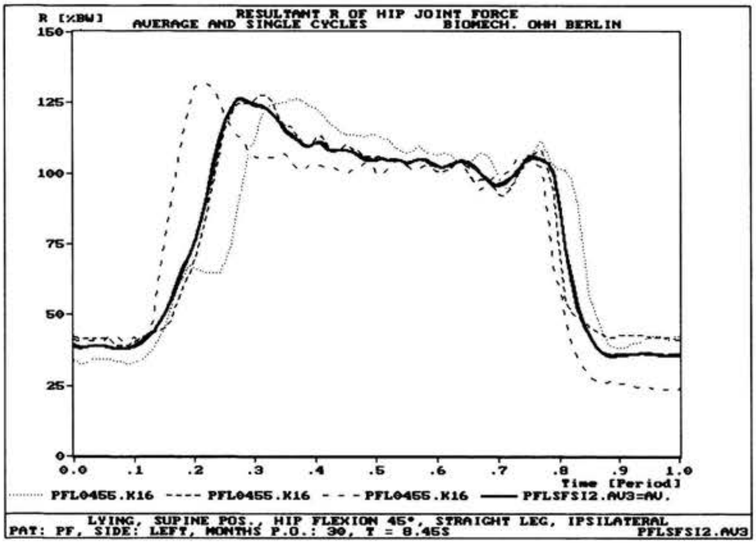

FIGURE 6. Lifting straight leg.
Resultant forces from 2 trials of 1 patient (thin) and their average (thick). The subject lies in supine position, lifts the straight leg by 45° and finally lies relaxed again.

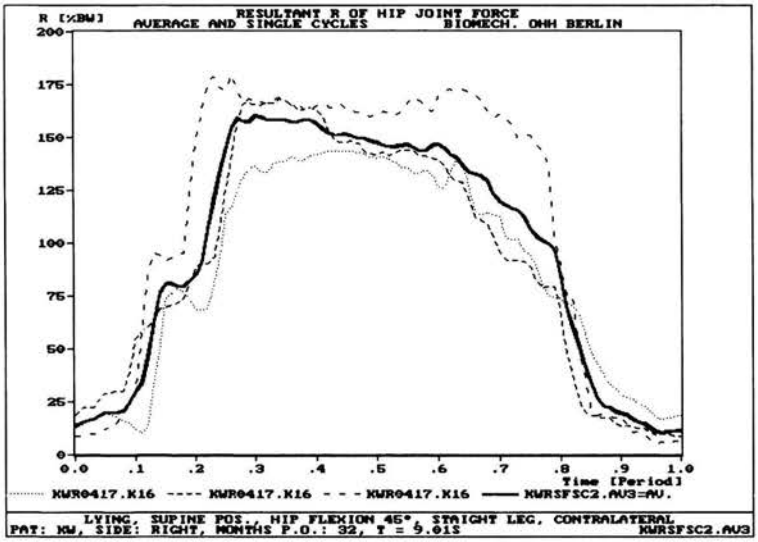

FIGURE 7. Lifting contralateral straight leg.
Resultant forces from 3 trials of 1 patient (thin) and their average (thick). The subject is lying in supine position and lifts the *contralateral* straight leg by 45°.

hip joints have to be stabilized by muscles when lifting one leg. ***Lifting the pelvis*** only slightly while lying on the back typically loads the joint with 112% BW and with 196% BW in a single subject (Fig. 8). The contact forces can even reach 284% BW when lifting the pelvis very high.

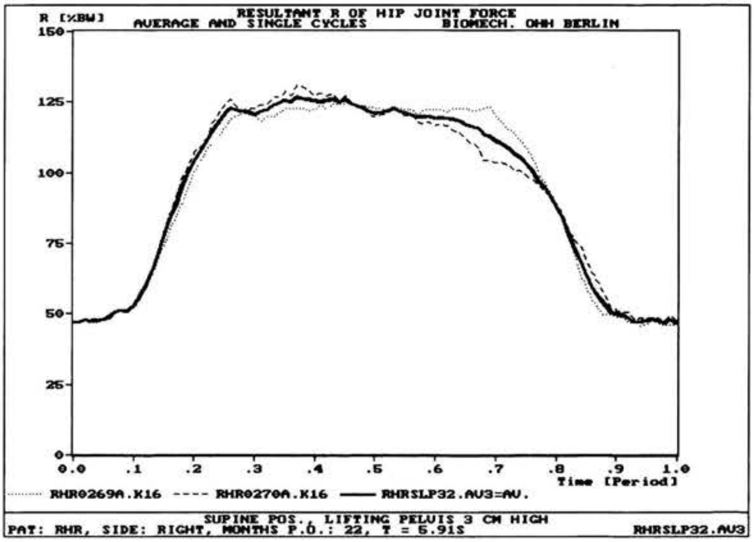

FIGURE 8. Lifting pelvis slightly in supine position.
Resultant forces from 2 trials of 1 patient (thin) and their average (thick). The subject is lying in supine position and lifts the pelvis 3 cm high.

The use of ***crutches*** is less effective than assumed by many clinicians. The load-reducing effect shows high inter-individual variation (Bergmann et al. 1992) but elderly patients cannot be expected to achieve a more than 30% reduction of contact forces and only during the first postoperative weeks.

Of particular interest were two measurements taken during ***stumbling*** (Bergmann et al. 1993). One of the patients missed the first step when going upstairs and the other stumbled over an electric cord while walking. Neither fell, but both took some quick steps to prevent falling. Peak forces reached 720% BW in the first case and 870% BW in the second. These were by far the highest hip joint loads observed for any activity.

The effect of ***shoes*** and ***floor materials*** on the magnitudes of hip contact forces was extensively investigated in one of the patients (Bergmann et al. 1995). The subject walked on a treadmill at a pace of 3 km/h with 14 different shoes of various designs and materials, from very soft tennis shoes to hard hiking boots, and the peak contact forces were compared to those observed when walking barefoot. Loads with and without shoes proved to be nearly the same, with some shoes they even increased slightly. No systematic relationship was found between the shoe design and force magnitudes. The torsional moment was markedly higher with most shoes. The influence of floor properties on the contact forces was investigated while the subject first walked barefoot or with shoes on hard ground and then on 2 and 4 cm thick gymnastic mats. Loads were lowest when walking on hard ground and increased on soft floor surfaces.

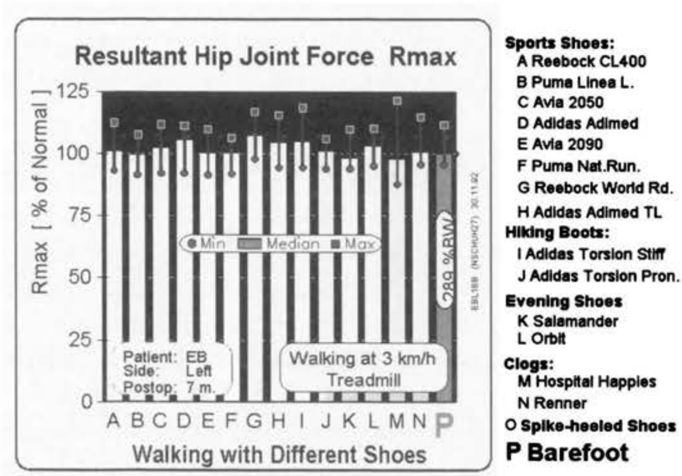

FIGURE 9. Walking with different shoes.
The peak hip contact forces when walking barefoot (right) were compared to those when wearing different shoes. None of the shoes lowered the joint load significantly. (With permission from Bergmann et al. 1995b).

The hip joint load was typically very low when using a stationary ***bicycle***. At a power of 100 watts and a speed of 60 rpm, the contact force remained below 115% BW with an average of 95% BW. Even at 160 watts, the average forces only reached 135% BW when cycling at a constant speed. The torsional moments acting to turn the implants backwards were also lower during cycling than walking.

One potential problem of hip arthroplasty caused by the head and cup materials is the ***frictional heating*** of implants (Bergmann et al. 2001a,b). Depending on the friction properties, the head and cup of the implant will warm up during long-lasting activities like walking or jogging. In five patients with aluminum ceramic implant heads and polyethylene cups, the highest temperature after an hour of walking was 43.1^{circ}C. However, this was measured in the most lightweight subject of the group; others with a 50% higher body weight had values as low as 40^{circ}C. Heating up of hip implants is a slow process. It takes about one hour of walking until the final temperature is reached, and the time required for cooling down the prosthesis is in the same range. Comparing the thermal effects of walking with those of cycling at the same metabolic power but much lower hip contact forces, we found that the final joint temperatures remain lower during cycling.

4. Discussion

It is nearly impossible to totally unload the hip joint. The contact force only remains below 30 or 40% BW during lying or completely passive sitting. Even lifting the pelvis slightly, as required when using a bedpan, already causes joint forces higher than the body weight. .

Even if a subject tries to keep the hip contact forces low, as in the case of cementless implants during the first postoperative months or when implant loosening is suspected, routine activities like walking or going upstairs can hardly be avoided. Preventing excessive hip joint loading primarily requires the patient to walk with caution and avoid stumbling. Crutches can only really reduce joint forces if strongly loaded, which is not achievable for most elderly patients. However, walking with crutches may help to train the patient's gait shortly after surgery.

When standing on two legs, only about 33% BW would be expected in each hip joint, because they carry an equal distribution of the upper body weight. Higher values are probably required for balancing an unstable position. This assumption is supported by the observation that the contact forces decrease slightly with increasing foot distance in most subjects.

Safe walking seems to be much more important than the use of crutches in the early postoperative phase. The principle behind cementless implants is for the bone to grow into the porous surface of the metallic interfaces and to thus increase fixation stability in the initial phase after arthroplasty. We suspect that extreme joint loads during stumbling may prevent or delay bone formation around the fixation surfaces.

The relatively high forces immediately before ground contact during walking are required to reverse the angular movement of the leg and decelerate the foot to achieve a soft heel contact. Force peaks transmitted from the floor to the foot at the instant of heel strike are always damped on their way up to the hip. Such damping can be achieved by the soft heel pad, by adapted movement of the lower limb segments and also by the femur, which bends under load due to its curved shape. This suggests that soft shoe materials or visco-elastic insoles, for example, are not suitable for lowering the peak loads acting at the hip joint. This is supported by the observation that heel strike happens too early to influence the maximum joint force, which occurs much later in the gait cycle. The fact that shoes often slightly increase the force magnitudes at the hip is most probably explained by the stability of gait, which seems to be best when walking barefoot on hard ground.

The use of a stationary bicycle can be recommended for general fitness training because the joint loads remain far below those during walking if cycling is not done at extreme power. Our data suggest that cycling does not involve a higher risk of implant loosening than other common activities.

Differences in the heating of hip implants during walking are most probably due to variation in the frictional properties and the volume of synovial fluid in the joint. The fact that these properties can hardly be assessed in patients prevents an individual estimation of the potential risk of thermally induced implant loosening. The general use of low-friction material combinations for the head and cup should therefore be a goal in developing future materials. The one case enabling a comparison of the final joint temperatures of ceramic and polyethylene cups articulating against a ceramic ball demonstrated that low-friction material combinations help to keep the implant temperatures low. Thus we would not suggest using a metal/polyethylene combination for the head/cup of hip implants because of higher friction and a possibly increased risk of thermally induced implant loosening, particularly in combination with frequent long walking periods. As expected, the comparison between walking and cycling clearly showed that higher contact forces cause higher implant temperatures after long-lasting activities. A high body weight is therefore disadvantageous. Although fast jogging was not investigated in our patients, we would not suggest this activity for patients with hip implants.

References

1. G. BERGMANN., F. GRAICHEN, J. SIRAKY, H. JENDRZYNSKI, A. ROHLMANN, Multichannel strain gauge telemetry for orthopaedic implants, *Journal of Biomechanics*, Vol.21, pp.169-176, 1988
2. G. BERGMANN, F. GRAICHEN, A. ROHLMANN, Load reduction of hip joint implants by forearm crutches, in: *Proceedings of the 8th Meeting of the European Society of Biomechanics*, p.300, 1992.
3. G. BERGMANN, F. GRAICHEN, A. ROHLMANN, Hip joint forces during walking and running, measured in two patients, *Journal of Biomechanics*, Vol.26, pp.969-990, 1993.
4. G. BERGMANN, F. GRAICHEN, A. ROHLMANN, Is staircase walking a risk for the fixation of hip implants?, *Journal of Biomechanics*, Vol.28, pp.535-553, 1995a.
5. G. BERGMNANN, H. KNIGGENDORF, F. GRAICHEN, A. ROHLMANN, Influence of shoes and heel strike on the loading of hip implants, *Journal of Biomechanics*, Vol.28, pp.817-827, 1995b.
6. G. BERGMANN, F. GRAICHEN, A. ROHLMANN, Hip joint forces during load carrying, *Clinical Orthopaedics and Related Research*, Vol.335, pp.190-201, 1997.

7. G. BERGMANN, F. GRAICHEN, A. ROHLMANN, N. VERDONSCHOT, G.H. VAN LENTHE, Frictional heating of total hip implants, Part 1: measurements in patients, *Journal of Biomechanics*, Vol.34, pp.421-428, 2001a.

8. G. BERGMANN, F. GRAICHEN, A. ROHLMANN, N. VERDONSCHOT, G.H. VAN LENTHE, Frictional heating of total hip implants, Part 2: finite element study, *Journal of Biomechanics*, Vol.34, pp.429-435, 2001b.

9. G. BERGMANN, G. DEURETZBACHER, M. HELLER, F. GRAICHEN, A. ROHLMANN, J. STRAUSS, G.N. DUDA, Hip contact forces and gait patterns from routine activities, *Journal of Biomechanics*, Vol.34, pp.859-871, 2001c.

10. G. BERGMANN (ed.) *HIP98 – Loading of the hip joint*, Free University of Berlin, Compact disc, ISBN 3980784800, 2001d.

11. G. BERGMANN, F. GRAICHEN, A. ROHLMANN, M.M. MORLOCK, M.O. HELLER, G.N. DUDA, Realistic test scenarios for hip implants, in: *Proceedings of the International Conference 'Engineers & Surgeons – Joined at the Hip'*, Abstract C601/029/2002, Institute of the Mechanical Engineers, London 2002.

12. R.A. BRAND, D.R. PEDERSEN, D.T. DAVY, G.M. KOTZAR, K.G. HEIPLE, V.M. GOLDBERG, Comparison of hip force calculations and measurements in the same patient, *Journal of Arthroplasty*, Vol.9, pp.45-51, 1994.

13. D.W. BURKE, D.O. O'CONNOR, E.B. ZALENSKI, H.E. RUBASH, W.H. HARRIS, Biomechanics of femoral total hip components in simulated stair climbing, in: *Proceedings of the 34th Annual Meeting of the Orthopaedic Research Society*, p.345, Atlanta 1988.

14. R.D. CROWNINSHIELD, R.C. JOHNSTON, J.G. ANDREWS, R.A. BRAND, A biomechanical investigation of the human hip. *Journal of Biomechanics*, Vol.11, pp.75-85, 1978.

15. F. GRAICHEN, G. BERGMANN, Telemetrisches Übertragungssystem zur in vivo-Messung der Hüftgelenkkraft mit instrumentierten Prothesen, *Biomedizinische Technik*, Vol.33, pp.305-312, 1988.

16. F. GRAICHEN, G. BERGMANN, A. ROHLMANN, Hip endoprosthesis for in vivo measurement of joint force and temperature, *Journal of Biomechanics*, Vol.32, pp.1113-1117, 1999.

17. M.O. HELLER, G. BERGMANN, G. DEURETZBACHER, L. DÜRSELEN, M. POHL, L. CLAES, N.P. HAAS, G.N. DUDA, Musculo-skeletal loading conditions at the hip during walking and stair climbing, *Journal of Biomechanics*, Vol.34, pp.883-893, 2001.

18. D. NUNN, M.A.R. FREEMAN, K.E. TANNER, W. BONFIELD, Torsional stability of the femoral component of hip arthroplasty, *Journal of Bone and Joint Surgery*, Vol.71B, pp.452-455, 1989.

19. F. PAUWELS, Der Schenkelhalsbruch. Ein mechanisches Problem. *Beitr. Orthop. Chir.*, Vol.63, p.1, 1935.

20. F. PAUWELS, *Atlas zur Biomechanik der gesunden und kranken Hüfte*, Springer, Berlin 1973.

21. B.W. STANSFIELD, A.C. NICOL, J.P. PAUL, I.G. KELLY, F. GRAICHEN, G. BERGMANN, Direct comparison of calculated hip joint contact forces with those measured using instrumented implants. An avaluation of a three-dimensional mathematical model of the lower limb, *Journal of Biomechanics*, Vol.36, pp.929-936, 2003.

22. K.E. TANNER, W. BONFIELD, D. NUNN, M.A.R. FREEMAN, Rotational movement of femoral components of total hip replacements in response to an anteriorly applied load, *Engineering Medicine*, Vol.17, pp.127-129, 1988.