Heat Generated by Hip Resurfacing Prostheses: An in Vivo Pilot Study

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ABSTRACT: In order to determine the magnitude of temperature increases in resurfaced hips, temperature sensors were placed percutaneously in both hip joints of 12 volunteer patients who had 1 or both joints resurfaced. Temperature recordings were made with patients at rest (baseline) and after patients walked for 20 and 60 minutes. The hip resurfacing procedures were performed 12 to 36 months prior to this study using 9 different acetabular bearing surface components. At baseline (resting), a ceramic femoral prosthesis articulating with a poly-ether-ether-ketone (PEEK) acetabular prosthesis generated a temperature increase of 4°C compared to a normal contralateral hip. After 60 minutes of walking, a ceramic femoral prosthesis articulating with a polyurethane acetabular prosthesis generated a temperature increase of 5°C, whereas a ceramic femoral prosthesis articulating with a metal acetabular prosthesis generated a temperature increase of 6°C, a cobalt-chromium alloy femoral prosthesis on a polyethylene acetabular prosthesis generated a temperature increase of 7°C, and a cobalt-chromium alloy metal-on-metal prosthesis generated a temperature increase of 8°C. Resurfaced hips generate more heat than arthritic and normal hips, and arthritic hips generate more heat than normal hips. A resurfaced hip with a ceramic femoral and PEEK or polyurethane acetabulum generated less heat than a resurfaced hip of the same design using a cobalt-chromium femur and either cobalt-chromium, or polyethylene for the acetabulum. Frictional heat generated in a resurfaced hip is not immediately dissipated and may result in increased bearing surface wear. Extended periods of elevated temperature within joints may inhibit periarticular cell growth and perhaps contribute to bone resorption or component loosening over the long term.

KEY WORDS: hip resurfacing, heat, joint friction

I. INTRODUCTION

The synovial fluid of natural and prosthetic joints increases in temperature as a result of friction during walking and other activities.[1–4] Measuring the temperature of synovial fluid provides a method for estimating joint friction.[1,5–7] There is some published information about heat generation in total hip prostheses using polyethylene and ceramic bearing surfaces.[5–9] However, to our knowledge, there is no published information comparing increases in joint temperature in normal, diseased, and resurfaced hips. Also, there is currently no information about the heat generated by metal-on-metal, polyurethane, and poly-ether-ether-ketone (PEEK) acetabular bearing surfaces.

The amount of frictional heat that occurs in a prosthetic joint is a function of speed and duration of movement, applied load, quality of lubricant, and the presence of wear debris. It may also be a function of prosthetic design and bearing surface material.[5–9] The amount of heat that is produced by a resurfaced hip is influenced by a number of factors: the volume of synovial fluid, the separation of the femoral head and acetabulum, perfusion to the joint, friction between the femoral head and the acetabulum, fixation technique of the prosthetic components, and the thermal properties and size of the components.[5–9]

Generally, heat is well tolerated and the efficiency of synovial fluid in lubricating both natural and prosthetic joints permits many years of successful use for most patients. However, excessive heat can accelerate wear and oxidative degradation of high-density polyethylene bearing surfaces and, possibly, other bearing materials.[10] It has been shown that a temperature increase within the joint of as little as 6°C can cause cell death, fibrous tissue formation and, possibly, periprosthetic pain or prosthetic loosening.[2,4,11]

Previously, we evaluated the amount of heat generated in replaced knees.[12] The present study was conducted to determine the amount of heat produced in resurfaced, arthritic, and normal hips and, secondarily, to compare and quantify the heat generated by prostheses with different bearing surfaces.

II. MATERIALS AND METHODS

This study was approved by the Institutional Review Board of Providence Hospital (Seattle) Twelve patients volunteered to participate in measuring the temperature of the synovial fluid in their hip joints and all patients provided written, informed consent. Seven patients (14 hips) had undergone bilateral hip resurfacing procedures and 5 of the 7 had a different prosthesis on each side (1 patient had bilateral metal-on-metal resurfacing prostheses and 1 had bilateral ceramic-on-PEEK resurfacing prostheses), and 5 patients had unilateral resurfacing procedures; the contralateral hip was normal in 2 patients and arthritic in 3 patients. Patients ranged in age from 33 to 66 years (mean, 52 years). Six patients were women (mean age, 51 years; range, 35-63 years) and 6 were men (mean age, 53 years; range, 33-66 years). The mean weight of the women was 77 kg (range 65-88 kg) and 91 kg (range 80-105 kg) for the men.

All of the hip resurfacing procedures were performed by a single surgeon using the same technique.[13] The anterolateral approach was used for all resurfacing procedures and all patients were followed using the same postoperative management protocol. Nine different bearing surfaces were used: metal-on-metal, metal-on-polyethylene, metal-on-polyurethane, ceramic-on-metal, ceramic-on-polyethylene, ceramic-on-PEEK, ceramic-on-polyurethane, ceramic-on-articular cartilage, and metal-on-articular cartilage. The hemiarthroplasty procedures had been performed for osteonecrosis of the femoral head with an intact acetabular cartilaginous surface, based on magnetic resonance imaging (MRI) and radiographic evaluation. The hemiarthroplasty procedures were all on the femoral side. The total resurfacing procedures were performed for osteoarthritis involving both the femoral head and acetabulum.

All of the femoral prostheses were of the same design (total articular replacement arthroplasty). The design utilizes a short, curved femoral stem. The metal prostheses for both the acetabulum (Vitallium®, Stryker, Mahwah, NJ) and the femur (BioPro, Inc., Port Huron, MI; DePuy, Warsaw, IN; Zimmer, Warsaw, IN) were cobalt-chromium alloy. The ceramic femoral component was a Zirconia-based ceramic (BioPro, Inc.) (Fig. 1). The polyethylene and polyurethane components were standard medical-grade conventional polymers. One polyethylene acetabular component was highly cross-linked. The femoral components ranged from 44 to 52 mm and the acetabular components ranged from 50 to 58 mm. The patients were from 12 to 36 months post-hip resurfacing at the time of this study, and all patients considered the result of their hip resurfacing procedure to be good or excellent.

A temperature sensor (400 Series, De Royal Co., Powell, TN) was placed percutaneously into both hip joints in an office examination room by the investigator using a direct anterior approach without local anesthesia. Return of synovial joint fluid into the introducer needle verified that the sensors were intracapsular. Manufacturer documentation stated that the sensor device produced readings accurate to within ± .02°C at 0°C to 50°C.

Following insertion of the temperature sensors, the patients walked approximately 20 minutes to the laboratory, where temperature measurements were obtained. The patients rested for 20 minutes before the initial baseline temperatures were re-

FIGURE 1. Metal, polyurethane, and polyethylene acetabular components with metal and ceramic femoral components.

corded. Next, each patient walked on a treadmill at a speed of 3 to 4 km/hr for 60 minutes. Temperature measurements were recorded after 20 and 60 minutes of walking. The sensors were removed after the 60-minute temperature measurements were completed. Therefore, the sensors were in place for approximately 1 hour. One hour was chosen as the end point because previous studies have found that steady-state temperatures are reached at that time.[1,5] All patients underwent follow-up examinations 2 and 6 weeks after the study, and annually thereafter.

III. RESULTS

None of the patients reported substantial pain during the testing and there were no complications associated with the insertion and removal of the temperature probes. Table 1 shows the mean intracapsular temperatures of normal and resurfaced hips; Table 2 shows temperature increases according to implant type. Normal hips (contralateral hip resurfaced) had a mean temperature increase of 1°C after 20 minutes of walking, and 2°C after 60 minutes of walking. Arthritic hips (contralateral hip resurfaced) had normal temperatures at rest, but temperatures increased by 2°C after 20 minutes and by 3°C after 60 minutes of walking. There were 3 patients with arthritic hips and 2 patients with normal hips (on 1 side), and each had the same increase in synovial fluid temperature with walking. The temperature of the synovial fluid increased by 4^{0}C with a metal hemiarthroplasty and 3^{0}C with ceramic hemiarthroplasty procedure, after 60 minutes of walking (Table 2).

At baseline, the temperature within resurfaced hips of every type was 1°C to 2°C higher than in a normal hip. In the patients with bilateral, metal-on-metal, and ceramic-on-PEEK resurfacing implants, the temperature increase was the same in both hips at the 20-minute and 60-minute temperature measurements. All the other prosthesis

TABLE 1. Mean Hip Joint Synovial Fluid Temperature Increases (°C) With Activity in Normal, Arthritic, and Resurfaced Hips

Hip Type (n)	Baseline	20 min.	60 min.
Normal (2)	36	1	2
Arthritic (3)	37	2	3
Resurfaced (19)	38	5	6

TABLE 2. Mean Hip Joint Synovial Fluid Temperature Increases (°C) With Walking by Prosthetic Type

Type of Prosthesis (n)	Baseline °C	20 min.	60 min.
Metal hemi (1)	38	2	4
Ceramic hemi (2)	38	2	3
Metal-on-metal (3)	39	8	8
Ceramic-on-polyethylene (2)	38	5	7
Metal-on-polyethylene (3)	38	6	7
Metal-on-polyurethane (2)	38	5	6
Ceramic-on-polyurethane (2)	37	4	5
Ceramic-on-metal (2)	38	5	6
Ceramic-on-PEEK (2)	37	4	4

materials showed an increase in the synovial fluid temperature from the 20- to the 60-minute measurement. The baseline temperature was higher for metal-on-metal by 1°C to 2^0C in comparison with the other bearing surfaces. Table 3 shows temperature data for all 12 patients.

IV. DISCUSSION

The main goals of hip arthroplasty are to reduce pain and increase function.[14] Sir John Charnley named his procedure "low-friction arthroplasty," indicating the importance of reducing friction in hip replacement.[15] There does not appear to be any previously published information concerning frictional heat generated in resurfaced hips, although heat generation has been documented in replaced knees.[12]

This study found that the baseline/resting synovial fluid temperature in a resurfaced hip is higher than the resting temperature within a normal, non-diseased joint. In addition, the temperature of the synovial fluid increases with walking and the amount of heat generated varies among the bearing surfaces used. The combination of a cobalt-chromium femoral head articulating with a polyethylene acetabular component generated more heat than a ceramic femoral head articulating with a polyethylene acetabulum. A ceramic femoral prosthesis articulating with a PEEK acetabular prosthesis generated the least amount of heat, and a metal-on-metal articulation generated the greatest amount heat. This study also found that arthritic hips generate more heat with walking than normal hips, but not as much heat as resurfaced hips. Hip resurfacing has been more successful using metal-on-metal components, but concerns remain about metal-on-metal articulations.[16,17]

During strenuous exercise, body temperature increases as the heat generated exceeds the body's ability to dissipate the heat.[5–7] Both *in vitro* and *in vivo* studies of total hip replacement prostheses suggest that the heat generated is not dissipated effectively and that the temperature increases with use. In an *in vitro* study,[7] a 32-mm cobalt-chromium alloy femoral head articulating with a polyethylene socket elevated the synovial fluid by 6°C to 7°C. A 32-mm aluminum oxide (ceramic) head articulating with polyethylene elevated the synovial fluid by 4°C to 7°C. The increase for an aluminum hydroxide femoral head articulating with an aluminum hydroxide socket was 3°C to 4°C. The synovial fluid in a normal hip (*in vivo*) increases by up to 2.5°C with activity.[3] However, it can be difficult to apply the findings of *in vitro* hip simulator studies into the human hip. *In vitro* studies use calf serum rather than synovial fluid, and a simulated joint capsule rather than natural tissue. Also, the

TABLE 3. Hip Joint Synovial Fluid Temperatures (°C) at Baseline and With Walking by Patient and Prosthetic Type

Patient	R Hip	L Hip	Baseline R/L	20 min R/L	60 min R/L
1	Normal	Metal hemi	36/37	37/39	38/41
2	Metal-on-metal	Degenerative joint disease	39/37	47/39	47/40
3	Ceramic hemi	Normal	38/37	39/38	41/39
4	Metal-on-polyethylene	Ceramic hemi	38/37	44/39	45/39
5	Degenerative joint disease	Metal-on-polyurethane	37/38	39/43	40/44
6	Degenerative joint disease	Metal-on-polyurethane	37/37	39/41	40/42
7	Ceramic-on-polyurethane	Metal-on-polyethylene	37/38	41/44	42/45
8	Ceramic-on-polyethylene	Ceramic-on-polyurethane	38/37	43/41	44/42
9	Ceramic-on-polyethylene	Ceramic-on-metal	38/38	43/43	45/44
10	Ceramic-on-metal	Metal-on-polyethylene	38/38	43/44	44/45
11	Metal-on-metal	Metal-on-metal	38/38	46/46	46/46
12	Ceramic-on-PEEK	Ceramic-on-PEEK	37/37	41/41/	41/41

thermal conditions for heat dissipation may not approximate the conditions in either a normal or resurfaced hip.[5–7] It is possible that after hip resurfacing surgery, tissues surrounding the hip joint become less efficient in dissipating heat generated within the joint.

In an *in vivo* investigation, Bergmann et al.[8] studied 7 patients with 5 instrumented hip prostheses using telemetry transmission data. They measured temperatures from the prosthesis directly rather than from the synovial fluid. The peak temperature in the femoral head was 43.1°C and varied considerably between the individual patients. One patient had 1 hip replaced with a ceramic femoral head articulating with a polyethylene acetabular prosthesis and the other hip replaced with a ceramic femoral head articulating with a ceramic acetabular prosthesis. The all-ceramic articulation had a peak temperature 1.8°C lower than the ceramic-on-polyethylene hip. The temperature of the synovial fluid was 2°C to 3°C higher than the temperature of the prosthetic femoral head. Another study by the same investigators found that a metal-on-metal total hip prosthesis generated more heat than a metal-on-polyethylene prosthesis.[9]

A sustained temperature increase of 6°C can induce fibrous tissue formation and periprosthetic bone resorption.[2,4,11] This can lead to prosthetic loosening and pain. An increase in the temperature of synovial fluid of 5°C caused precipitation of the lubricating proteins in simulation tests.[1] It is possible that a vicious cycle could be triggered beyond this temperature, as the lubricating properties of synovial fluid may vary among individuals. The temperature increases with some hip replacement and resurfacing procedures are enough to produce polyethylene wear, creep, and oxidative degradation.[18] However, it is not known how long the increased temperature must be maintained to produce damage in bone cells. Moritz and Henriques[19] reported that epithelial necrosis occurs 30 seconds after exposure to 55°C, and 5 hours after exposure to 45°C. Lundskog[20] noted bone cell necrosis after 30 seconds exposure at 50°C. However, due to the paucity of studies regarding thermal damage causing bone necrosis, temperature data from these two studies should be interpreted with caution.

There is a great deal of individual variation in temperature of synovial fluid within joints.[8,9] Under peak loading conditions, the temperature of synovial fluid can be elevated as much as 10°C. Ceramic and metal materials conduct the temperature away from the joint to the supporting bone much more effectively than polyethylene. This increases the temperature in the adjacent bone but also effectively lowers the synovial fluid temperature. Thus, a metal-backed acetabular component will reduce the temperature within the joint by 0.80°C as compared to a non-metal-backed, cemented prosthesis. A cobalt-chromium femoral stem will reduce the temperature within the joint by 0.70°C as compared to titanium, because it is a better temperature conductor. However, this is of less importance in resurfacing implants than in total hip implants because of the smaller amount of metal that is used.[8,9] Recently, Li et al.[21] developed a numerical finite element method to simulate bone cement polymerization, heat generation, and conduction in hip implants. Their method enables researchers to predict temperature distribution and polymerization reaction and makes it possible to quantitatively simulate the thermal behavior of bone-cement-prosthesis. This has potential application in improving implant designs before clinical trials occur.

The perfusion to the joint capsule and surrounding tissue can vary with temperature, the method of surgery, tissue healing, and the condition of the patient. Increased tissue perfusion can decrease the joint temperature by 0.60°C.[5–7] Patient size, activity type, and activity level also can result in individual temperature variation; more heat will be generated with jogging and less with a low-impact activity such as cycling. The volume of synovial fluid can also influence the heat transfer to surrounding tissues. Increased joint fluid can decrease the synovial fluid temperature by 1.5°C.[3]

The extent and manner of how frictional heat generated by hip prostheses affects wear, loosening, and pain are unknown. The temperature in metal-on-metal prostheses may change over time with metallosis or a lymphocytic reaction to metal wear debris.[16,17] Although it seems intuitive that younger, more active patients would generate more heat within the joints, this was not established by this study. Interestingly, patients rarely mention that their replaced hip(s) feel(s) warm; however, patients with knee replacements sometimes mention that their replaced knees feel warm.[2,12,22]

The present study found that resurfacing hip prostheses using a PEEK acetabulum (Fig. 2) generated the least amount of increased heat, followed by polyurethane, possibly because both PEEK and polyurethane are more hydrophilic than polyethylene.[12,13,22–24] The wear rate of both PEEK and polyurethane is less than that of conventional polyethylene and may be comparable to cross-linked polyethylene.[22,25] The present study did not find a difference in heat generation between cross-linked and conventional polyethylene. Polyurethane wear debris contains few submicron particles and may

FIGURE 2. A PEEK acetabular component.

also result in less bone loss from osteolysis.[22–24] Both PEEK and polyurethane have been used as hemiarthroplasty material against normal acetabular cartilage and produce limited wear on both the cartilage and polymer.[26] Polymer-on-polymer wear couples using polyethylene and polyacetal have been used experimentally in total knee replacement, but not for hip replacement or resurfacing.[27]

Polyurethane and PEEK have not been used widely for joint replacement prostheses,[24,28] although polyurethanes are used for artificial disc implants.[29] PEEK has been used for total hip replacement in clinical trials with promising results.[30] Polyurethane was first used for actetabular resurfacing in 1960 and provided excellent pain relief, but it did not wear well and was abandoned in favor of metal and, later, polyethylene.[13,31]

There are limitations to the present study. First, only a small number of patients were evaluated. However, given the invasive nature, although minimal, of the study, greater numbers of patients would likely be difficult to enroll. Also, patients were observed for a limited amount of time and only for a limited range of activity. A much larger number of observations would be necessary to be able to conclude whether the amount of heat generated is clinically relevant to prosthesis survivorship or joint function. A second limitation may be that this study is a comparison study that includes a single cohort. However, comparison studies within patients (paired) and not between patients have the advantages that fewer patients are required, confounding variables are controlled, and patient size, activity level, and size of the prosthesis all remain the same. Tissue perfusion, hip separation, and the amount of synovial fluid would also be expected to be similar in both hips of the same patient. Also, traditional parallel group trials may have bias, ie, the characteristics of synovial fluid may vary between different individuals. We used the same femoral design for each patient, so the only variable was the acetabular bearing surface. Because metal-on-metal and ceramic-on-metal implants have a wearing-in phase, only patients who were between 12 and 36 months post-resurfacing were included for temperature measurements.[16]

In conclusion, this study documented that the temperature of the synovial fluid in the hip increases with walking and that different hip resurfacing bearing surfaces generate different amounts of frictional heating. The combination used most commonly for a total hip replacement is a cobalt-chromium femoral component articulating with a polyethylene acetabular component. When used for hip resurfacing, however, the cobalt-chromium on polyethylene combination generated more heat than a ceramic femoral prosthesis articulating with a PEEK or polyurethane acetabular component. Currently, a cobalt-chromium metal-on-metal articulation is the bearing used most commonly for hip resurfacing, but this generates more heat than the other components studied. Further studies with a larger number of observations are needed to determine if the differences in temperature documented by this study will lead to clinically measurable and meaningful consequences in wear and/or prosthetic loosening over time.

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