Different thermal conductivity in drilling of cemented compared with cementless hip prostheses in the treatment of periprosthetic fractures of the proximal femur—an experimental biomechanical analysis

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Abstract

Purpose The purpose of this study was to evaluate the different temperature levels whilst drilling cemented and cementless hip prostheses implanted in bovine femora, and to evaluate the insulating function of the cement layer.

Methods Standard hip prostheses were implanted in bovine donor diaphyses, with or without a cement layer. Drilling was then performed using high-performance-cutting drills with a reinforced core, a drilling diameter of 5.5 mm and cooling channels through the tip of the drill for constantly applied internal cooling solution. An open type cooling model was used in this setup. Temperature was continuously measured by seven thermocouples placed around the borehole. Thermographic scans were also performed during drilling.

Results At the cemented implant surface, the temperature never surpassed 24.7 °C when constantly applied internal cooling was used. Without the insulating cement layer (i.e. during drilling of the cementless bone–prosthesis construct), the temperature increased to 47 °C.

Conclusion Constantly applied internal cooling can avoid structural bone and soft tissue damage during drilling procedures. With a cement layer, the temperatures only increased to non-damaging levels. The results could be useful in the

treatment of periprosthetic fractures with intraprosthetic implant fixation.

Introduction

The traditional approach to the treatment of periprosthetic fractures of the proximal femur with a stable implanted hip prosthesis involves the use of unicortical locking plates at the level of the prosthesis [1–4].

To achieve higher primary stability we evaluated an intraprosthetic fixation system that still uses a locking plate system [5]. After developing a usable drill, which could be connected to hand drills routinely available in trauma surgery, our experimental studies showed the necessity for a different, more effective, cooling system for intraprosthetic drilling of implant and cement structures [6].

Drilling of every material is always associated with the conversion of mechanical energy into thermal energy. Therefore, temperatures rise above normal physiological levels of bone and surrounding soft tissue transiently [7–11].

Earlier studies have investigated the temperature increase whilst drilling human bone tissue, and how it relates to the amount of damage to surrounding vital tissue [12–14]. Lundskog et al. described a significant decrease in vital bone tissue starting at a temperature of 50 °C and exposure time of 30 seconds [10]. Augustin et al. reported in their review of cortical drilling a threshold level for bone survival during implant site preparation of 47 °C with a drilling time below one minute [8]. Externally applied cooling with saline solution, an increase in applied feed-rate and increased rotational speed of the drill led to a smaller increase in temperature at the bone surface [15–18]. Stability of osteosynthesis was reduced with thermally damaged bone

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tissue possibly due to reabsorption phenomena around the screws [19].

When drilling solid materials, such as standard hip prostheses made from titanium or steel, the machining temperature can rise up to 520 °C (steel) or 604 °C (titanium). Even with externally applied cooling, an increase in feed-rate and greater rotational speed, drilling temperatures still increase to tissue-damaging levels. Increases in drilling time can also lead to tissue-damaging temperatures.

Further investigations have shown a decrease in temperatures to where vital tissue is preserved, by using step drills and an internally applied cooling system [9].

Our hypothesis for the experimental trials presented below was that, when using HPC drills (high performance cutting) and constantly applied internal cooling in an open type model, drilling into prostheses would not lead to tissue damaging temperatures, regardless of the prosthesis fixation technique (i.e. with or without bone cement).

Second, we expected the temperature increase when drilling cemented prostheses to be moderate compared to that found when drilling cementless prostheses because the cement layer is insulating.

Materials and methods

Drilling machine

HPC drills with an open type internal cooling system were used (Fig. 1). A rotational speed of $n=500 \text{ min}^{-1}$ and a force of 400 N were applied to the 5.5-mm diameter HPC-drill during drilling. The feed-rate was applied by hand to the drill. The drill and prepared specimen were mounted on a stationary testing device. The HPC driller ($\sigma=140^\circ$; 5.5-mm diameter; Perschmann®, Munich, Germany) with reinforced core and a high level of concentricity was used. The cooling material transfer occurs via spiral channels through the tip of the drill (diameter 0.4 mm).

Specimen preparation

For testing of temperature increases during intraprosthetic drilling of cemented and cementless hip prostheses, standard hip prostheses (Aesculap type Excia Isotan with plasmapore coating, base material TiA16V4; B. Braun®, Melsungen, Germany)

were implanted into bone tissue (bovine femoral diaphyses). Implantation was secured either via a press-fit procedure or using standard bone cement (Biomet, Warsaw, IN, USA).

Temperature testing

Seven thermocouples were placed on the implant/bone tissue construction to measure temperature changes during drilling. Of these, four thermocouples were placed one millimetre from the drill hole at the surface of the implanted prosthesis to measure the changes in temperature at the interface between implant and bone tissue during drilling. The other three thermocouples were placed horizontally, spaced three millimetres apart, near the drilling hole to measure temperature changes at the interface between the drill and bone tissue (Fig. 2). Placement of thermocouples was marked with a laser and fitted manually. Saline solution at room temperature was used as the internally applied cooling solution; the volume flow was 14.1 ml/s, a lower volume flow was not tested in this study design.

After preparation, the drilling procedure was performed using the 5.5-mm HPC drill and a drilling time of ten seconds. The temperature trend was measured at the seven thermocouples. Additionally, thermographic scans were performed during drilling using a thermographic camera (Flir Thermacam, thermic sensitivity 0.02 k, Flir Systems®, Wilsonville, OR, USA) (Fig. 3). Thermographic scans were only used additionally for the purpose of assessing the coolant temperature.

Statistical analysis

To analyse the damage to bone tissue caused by increased temperatures, data can be fitted by regression to the Lundskog-formula: $T_o = T_d - 63.55 \cdot t_d - 0.0405$, where T_o is the bone tissue damaging temperature (°C), T_d is the drilling temperature (°C) and t_d is drilling time (s).

Using this regression model, thermal tissue damage is expected when T_o is positive, and no thermal damage is expected when T_o is negative [10].

Results

The use of internally cooled HPC drills effectively suppressed the rapid temperature increases at the surfaces of both the implants and bone. In particular, the temperature rise caused

Fig. 1 HPC (high performance cutting) drill with internally applied cooling system via cooling channels through the tip ($\sigma=140^\circ$; Perschmann)

Fig. 2 Arrangement of thermocouples and experimental drilling setup

by machining of the implant was effectively suppressed. At the surface of the cemented implant, the measured temperature remained below 24.7 °C (Fig. 4). Because of the larger thermal conductivity of bovine bone in our setup ($\lambda=0.3\text{--}0.6\text{ W/mK}$) compared to that of bone cement ($\lambda=0.19\text{ W/mK}$) the maximum temperature was reached more quickly at the implant, compared to the surrounding bone cement (Fig. 4). The thermal insulating function of bone cement did not delay the temperature increase at the implant. Within the bone cement, the increase in temperature during drilling was much slower. The machining process led to a rapid temperature increase, which was mostly confined to the bone tissue surface, because the insulating bone cement prevented direct transfer of heat between bone tissue and implant.

At the maximum drilling depth, the smaller thermal conductivity level of bone cement and titanium implant increased the time required for the temperature to reach a maximum.

Fig. 3 Temperature changes during drilling of cemented implants with an internally cooled HPC (high performance cutting) drill ($D=5.5\text{ mm}$)

Finally, the time of drilling is increased when drilling a cemented prosthesis; this means that the cooling coil extends further and can cover more tissue.

Using HPC drills with internal cooling application for intraprosthetic drilling in cemented titanium hip prostheses did not lead to any significant thermal damage to the surrounding tissue. All measured temperatures during the drilling procedure were far below the protein denaturation temperature.

Also, drilling in cementless hip prostheses made of titanium did not cause the temperatures of the tissues to exceed the osteonecrosis temperature. Special requirements included the use of an internally cooled HPC driller and the implantation of a titanium hip prosthesis.

With temperatures only increasing to a maximum of 24.7 °C during drilling cemented hip prosthesis thermally induced damage to the surrounding cement layer from drilling seems likely to be negligible, especially when considered in relation to the temperatures of between 46 °C and 124 °C, which arise during hardening of the cement as a result of exothermic polymerisation reactions.

Discussion

Analysing the measured temperatures at the drill-hole level and implant surface suggests that thermally caused damage to vital bone and surrounding tissue seems unlikely. Because of its low thermal conductivity, bone cement forms an insulating layer, and protects the more temperature-sensitive bone tissue below.

In contrast, when drilling cementless implants there is no insulating layer. This leads to an immediate heat flow into the surrounding bone tissue, and a more rapid rise in temperature. All thermocouples reached their maximum temperature (around 47 °C) at the same time; this maximum temperature was maintained for a short period (Fig. 5).

The reasons for the higher drilling temperatures when drilling cementless implants are, first, the absent insulating

Fig. 4 Temperature changes during drilling of cementless implants using an internally cooled HPC (high performance cutting) drill ($D=5.5$ mm)

cement layer and, second, the lack of covering bone cement to absorb heat, so the heat input to the bone tissue is higher.

The goal of our experimental biomechanical study was to evaluate whether the cement layer around human hip prostheses has any significant influence on the drilling temperature reached during intraprosthetic fixation.

The results we achieved showed no significant tissue damage when drilling either cemented or cementless hip prostheses. The temperature increase during drilling cemented solid materials seems to be much slower and lower compared to that found during drilling of cementless prostheses. This is because of the insulating cement layer, the prolonged temperature flow, the absorption of heat by the bone cement and the prolonged drilling time. However, by using internally cooled HPC drills, both cemented and cementless prostheses are suitable for intraprosthetic fracture fixation without causing serious damage to surrounding vital bone tissue.

Considering the increasing number of periprosthetic fractures of the proximal femur, we had the idea of achieving a higher level of primary stability by developing an intraprosthetic fixation system. During this development

process we realised that there were special problems to be resolved. First, we had to develop a drill strong enough to drill solid materials, such as hip prostheses. Second, we discovered that the rapid and massive increase in temperature during drilling caused immediate tissue damage. This meant that, third, we had to develop a special internally applied cooling system, and had to prove experimentally that our system was superior to common cooling methods, such as externally applied saline solution.

In particular with older, thinner cortical bone, we expected intraprosthetic fixation to produce better primary stability than the commonly used monocortical fixation.

For our experimental biomechanical program we used an HPC drill with a newly designed internal cooling system. Using this system, we were able to decrease the drilling temperature so that thermal damage to bone tissue did not occur.

The experimental design of this biomechanical study is limited, and further studies need to include work using the drilling machine and intraprosthetic fixation on paired human femora. Another concern is the possibility of weakening the prosthesis by drilling; this seems implausible but not impossible. Thus, further investigations are needed to evaluate the amount of damage intraprosthetic drilling causes to solid hip prostheses.

Fig. 5 Thermographic scan of drilling procedure

Conclusion

One crucial aspect of implementing a new method of intraprosthetic fixation for the treatment of periprosthetic fractures of the proximal femur was to control the temperature increase during drilling of the solid materials of the prosthesis. Using HPC drills with high feed forces and a specially developed internal cooling system, we were able to drill both cemented and cementless hip prostheses while avoiding temperature rises that might have caused severe damage to the surrounding tissue. Comparing the drilling of cemented hip prostheses with the drilling of cementless hip prostheses showed that the cement acted as an insulating

layer and protected the bone. Thus, the temperature only reached 24.7 °C during drilling of cemented prostheses. The lack of insulating cement layer meant that higher temperatures were reached during drilling of cementless prostheses. However, by using constantly applied internal cooling, the drilling temperature did not cause significant tissue damage.

Statement All authors have participated in the research presented. The article has not been submitted elsewhere. An IRB Approval is not necessary with this study design.

Conflicts of interest and source of funding For all authors none declared.

References

1. Rayan F, Haddad F (2010) Periprosthetic femoral fractures in total hip arthroplasty—a review. *Hip Int* 20(4):418–426
2. Lever JP, Zdero R, Nousiainen MT, Waddell JP, Schemitsch EH (2010) The biomechanical analysis of three plating fixation systems for periprosthetic femoral fracture near the tip of a total hip arthroplasty. *J Orthop Surg Res* 5:45. doi:10.1186/1749-799X-5-45
3. Muller M, Kaab M, Tohtz S, Haas NP, Perka C (2009) Periprosthetic femoral fractures: outcome after treatment with LISS internal fixation or stem replacement in 36 patients. *Acta Orthop Belg* 75(6):776–783
4. Tarnowski JR, Holck K (2008) Osteosynthesis of a periprosthetic fracture of the proximal femur with the distal femur LISS system. *Acta Orthop Belg* 74(1):125–127
5. Brand S, Klotz J, Hassel T, Petri M, Haasper C, Bach FW, Krettek C, Goesling T (2012) Intraprosthetic fixation techniques in the treatment of periprosthetic fractures—a biomechanical study. *World J Orthop* 3(10):162–166. doi:10.5312/wjo.v3.i10.162
6. Brand S, Klotz J, Petri M, Ettinger M, Hassel T, Krettek C, Goesling T, Bach FW (2013) Temperature control with internally applied cooling in solid material drilling: an experimental, biomechanical study. *Int Orthop*. Mar 20 [Epub ahead of print]. doi:10.1007/s00264-013-1850-4
7. Augustin G, Davila S, Udiljak T, Vadrina DS, Bagatin D (2009) Determination of spatial distribution of increase in bone temperature during drilling by infrared thermography: preliminary report. *Arch Orthop Trauma Surg* 129(5):703–709. doi:10.1007/s00402-008-0630-x
8. Augustin G, Davila S, Mihoci K, Udiljak T, Vadrina DS, Antabak A (2008) Thermal osteonecrosis and bone drilling parameters revisited. *Arch Orthop Trauma Surg* 128(1):71–77. doi:10.1007/s00402-007-0427-3
9. Augustin G, Davila S, Udiljak T, Staroveski T, Brezak D, Babic S (2012) Temperature changes during cortical bone drilling with a newly designed step drill and an internally cooled drill. *Int Orthop* 36(7):1449–1456. doi:10.1007/s00264-012-1491-z
10. Lundskog J (1972) Heat and bone tissue. An experimental investigation of the thermal properties of bone and threshold levels for thermal injury. *Scand J Plast Reconstr Surg* 9:1–80
11. Bertollo N, Milne HR, Ellis LP, Stephens PC, Gillies RM, Walsh WR (2010) A comparison of the thermal properties of 2- and 3-fluted drills and the effects on bone cell viability and screw pull-out strength in an ovine model. *Clin Biomech (Bristol, Avon)* 25(6):613–617. doi:10.1016/j.clinbiomech.2010.02.007
12. Rafel SS (1962) Temperature changes during high-speed drilling on bone. *J Oral Surg Anesth Hosp Dent Serv* 20:475–477
13. Wiggins KL, Malkin S (1976) Drilling of bone. *J Biomech* 9(9):553–559
14. Fuchsberger A (1987) Effect of temperature on compact bone in saw cutting in relation to the conditions of use. *Zentralbl Chir* 112(12):793–804
15. Matthews LS, Hirsch C (1972) Temperatures measured in human cortical bone when drilling. *J Bone Joint Surg Am* 54(2):297–308
16. Abouzgia MB, James DF (1995) Measurements of shaft speed while drilling through bone. *J Oral Maxillofac Surg* 53(11):1308–1315, discussion 1315–1306
17. Abouzgia MB, Symington JM (1996) Effect of drill speed on bone temperature. *Int J Oral Maxillofac Surg* 25(5):394–399
18. Reingewirtz Y, Szmukler-Moncler S, Senger B (1997) Influence of different parameters on bone heating and drilling time in implantology. *Clin Oral Implants Res* 8(3):189–197
19. Lavini FM, Brivio LR, Leso P (1994) Biomechanical factors in designing screws for the Orthofix system. *Clin Orthop Relat Res* 308:63–67