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CURRENT PROBLEM CASE

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The relevance of hydroxyapatite and spongious titanium coatings in fixation of cementless stems

An experimental comparative study in rat femur employing histological and microangiographic techniques

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Abstract Pure titanium rods plasma-spray coated with hydroxyapatite (HA) or porous titanium (Ti) of controlled roughness were implanted bilaterally in the distal femur of Sprague-Dawley rats to compare the extent of bone growth on the two types of coating. The relevance of other factors, like mechanical stability and biological adaptation of the bone to the insertion of a foreign body implant, were investigated in femora which were overreamed (absence of primary fit) or reamed without insertion of the rod. Continuous tetracycline labeling for the first 30 days and for the last 2 weeks in the 90-day group was performed; histological/histometric, fluorescence and microangiographic studies were carried out on serial sections of the implanted and control femora. In the group of stable implants, HA-coated rods showed 90% integration versus 53% with Ti-coated implants ($P < 0.001$); in overreamed implants neither surface bone growth nor endosteal fixation occurred, and both types of rods were surrounded by a thick layer of connective tissue. The study documented early adhesion of osteoblasts and direct deposition of bone matrix on the substrate, while on spongious titanium osteogenesis was observed only in proximity to the surface. Remodeling of the reactive, primary bone to mature, lamellar bone took the form of a capsule surrounding the implants and radial bridges connecting the latter to the endosteal surface. The number, height and thickness of these bridges appeared to be the factors determining implant stability, rather than the extent of the bony capsule on the perimeter of the implant. Integration was a function not only of mechanical conditions and surface geometry, but also of the biological response of the whole bone to changes in the vascularization pattern. The reported phenomena can be seen more easily in experimental models involving small rodents because of their fast bone turnover and revascularization, but it is expected that they take place, even at a lower speed, in clinical situations like cementless stems of total hip replacement.

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Introduction

Hydroxyapatite (HA) coatings have gained favour in recent times as an easier way to achieve fixation in bone of cementless implants. There is evidence that the bond between the HA substrate and bone matrix is chemical in nature [1–4] and that HA coating induces more bone formation inside the cavities of a porous surface [5–9].

Under weight-bearing conditions the osteoconductive HA coating has been shown to increase the amount of bone ingrowth, but did not affect the mineral apposition rate of bone, which had a lower mineral content at the implant interface [10–12]. However, the chemical and physical properties of the implant surface are not the only factors which determine the type and strength of bonding; mechanical stability and biological reactivity of bone also have a relevant role to play [13–15].

In studies of unloaded, stable implants in canine femoral condyles, HA coating eliminated the negative influence of non-interference-fit between bone and implant [16] and enhanced bone tissue growth across a defined gap of 1 mm in the osteopenic cancellous bone of the condyle compared with control cancellous bone of normal density [17].

The purpose of the present study was to compare the bone response to HA and spongious titanium coatings with controlled surface roughness in an experimental model involving different situations of primary fit stability.

The biological response to implant insertion was investigated by histological and microangiographic techniques.

Materials and methods

Twelve Sprague-Dawley male rats (Stefano Morini, S. Polo d'Enza, Reggio Emilia, Italy), weighing about 400 g, received an

Table 1 Experimental design (*HA* hydroxyapatite, *Ti* titanium)

Time interval	Press-fit implant		Non-interference-fit implant		Controls (no implant)	
15 days	rats: 2	right: HA left: Ti			rats: 2	right: reamed left: intact
90 days	rats: 5	right: HA left: Ti	rats: 5	right: HA left: Ti	rats: 2	right: reamed left: intact

Table 2 Roughness values of HA and Ti coatings of implant (in μm)

	Ra	Rz	Rmax	Rq	Rpm	Rpm	R3z	Rt
HA coatings	2.998	25.169	27.068	3.791	15.43	12.568	19.515	30.267
Ti coatings	9.931	69.537	85.625	12.534	36.2	34.462	54.509	85.725

HA-coated rod in the right femur and a spongious titanium-coated rod of the same diameter in the left femur.

Two rats were killed after a time interval of 15 days and used for a qualitative, non-morphometric study of the early bone reaction to the intramedullary implant. The remaining ten were divided into two groups: the first had a press-fit implant inserted, the second a non-interference-fit implant. A further four rats had the right femur reamed but without implant insertion and the opposite femur left intact: they served as controls for the qualitative, non-morphometric study (Table 1).

The implants consisted of commercial pure titanium rods (diameter 1.5 mm and length 5 cm) coated by either HA or spongious titanium layers, which were deposited by vacuum plasma spray technique [18]. With this technique, depending on the deposition parameters, it was possible to obtain HA coatings of high crystallinity (> 60%) without further post-deposition thermal treatments, which always lead to composition modification and a degradation of the mechanical characteristics [19]. The evaluation of HA crystallinity was carried out through spectra math analysis obtained by X-ray diffraction of the coatings sprayed on check plates. Moreover, the plasma spray technique produces titanium coatings uncontaminated by oxides and nitrides with a controlled thickness and surface morphology.

The roughness values of the coatings were evaluated with a laser profilometer and are reported in Table 2.

Under ether anaesthesia, the right and left knee joints were exposed through a lateral parapatellar incision. A hole was drilled in the intercondylar notch, and the medullary canal was manually reamed up to 1.5 mm in diameter. After irrigation with sterile saline, a probe was inserted to assess the correct length of the implant. It was shortened appropriately with sterile cutting nippers and then impacted inside the reamed medullary canal; a pusher was used for the last few millimeters so that the distal end of the rod projected beyond the plane of the articular cartilage.

For rats in the non-interference-fit groups, the medullary canal was reamed up to 1.8 mm and the implant inserted in the same way, but no hammering was necessary to insert the rod.

For control rats the right femur was reamed without inserting an implant and the opposite femur left intact.

Rats were housed two to a cage, with free access to water and food. Tetracycline 30 mg/kg was injected intraperitoneally every day for the first 30 days; the group of rats killed after 90 days received a further dosage of tetracycline for the last 15 days.

Four rats (2 implanted and 2 controls) were killed with an overdose of ether 15 days after the operation, and the remaining 12 (10 implanted and 2 controls) after 90 days (Table 1).

The abdominal aorta was cannulated and the vascular tree of the hind limb washed with heparinized saline solution collected through the caval vein, which was then ligated to increase the perfusion pressure of the Indian ink solution (50%) injected into the artery. Both femora were dissected free from soft tissues and fixed in neutral formalin (10%). After X-ray of the specimen on a Kodak mammographic film, 3-mm-thick sections, starting from the intercondylar notch, were cut with a low speed saw (Isomet, Buheler Ltd, Lake Bluff, Ill., USA). The cutting plane was perpendicular to

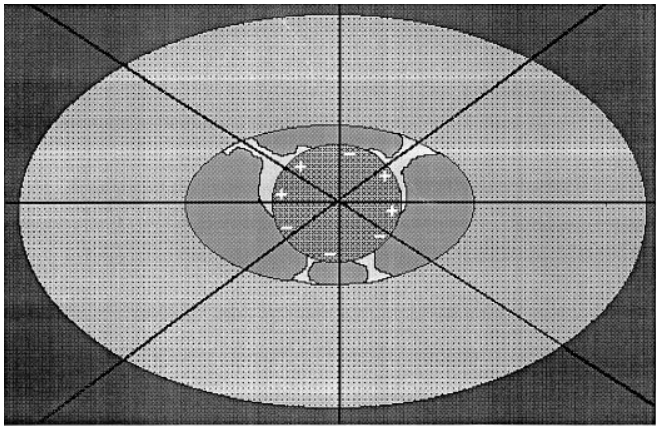


Fig. 1 Quantitative evaluation of implant integration in bone was carried out with a radial reticulum. Intersections of implant-bone interface are classified I^+ when there was contact between bone and implant surface without interposition of soft tissue. Integration percentage is expressed as the ratio of number of I^+ and the total number of intersections

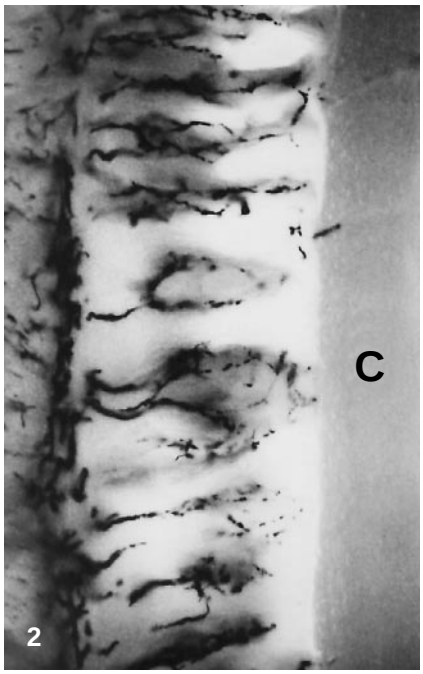
Fig. 2 Periosteal reaction 15 days after reaming (flurorescence and vessels injection, × 150). Hypertrophy of the periosteal vessels and development of radial branches. New bone labelled by tetracycline *(bright band)* has formed around these vessels at the periphery of the original cortex *(C)* ▶

Fig. 3 A, B Endosteal reaction 15 days after reaming (phase contrast, × 400). **A** Differentiation of osteoblasts and osteogenesis in proximity to spongious titanium coating. **B** Direct matrix apposition by osteoblasts on hydroxyapatite (HA)-coated rods

Fig. 4 Control femur reamed and not implanted after 90 days. Revascularization of the medullary canal is far advanced; reactive, endosteal bone has been completely resorbed, and the only sign of reaming is the devascularized area in the centre of the canal (haematoxylin & eosin, × 25)

Fig. 5 A–C Organization of endosteal, lamellar bone 90 days after implantation of HA-coated rod. **A** Remodeling of primary, endosteal bone formed thin bridges connecting the latter with the endosteal surface (haematoxylin & eosin, × 25). **B** Detail of radial bridges connecting HA-coated implant with the endosteal surface (haematoxylin & eosin, × 100). **C** Same field as in **B**, showing the lamellar organization of bone (polarized light, × 100)

Fig. 6 A, B Organization of endosteal bone 90 days after implantation of titanium (Ti)-coated rods. **A** Formation of radial bridges connecting the implant and endosteal surface. Revascularization of the medullary space is evident (haematoxylin & eosin, × 25). **B** Detail of bridge extension on the Ti-coated surface (phase contrast, × 250)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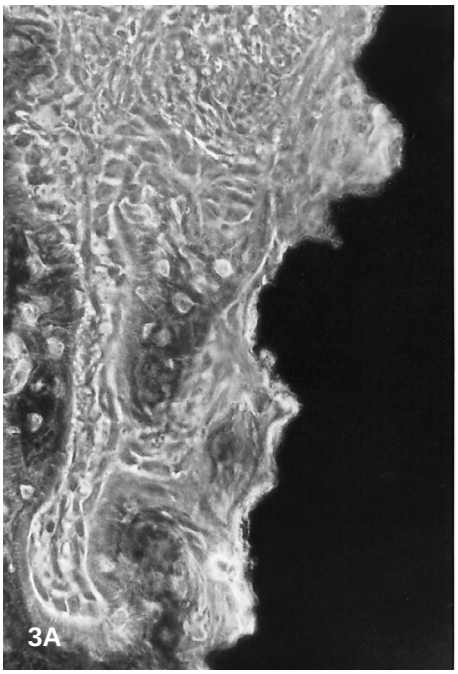
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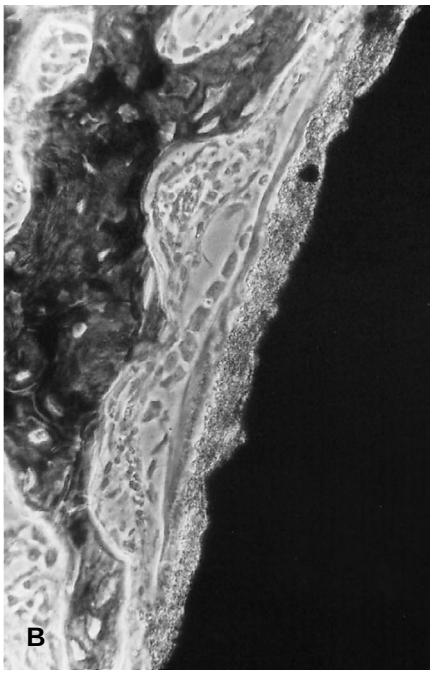
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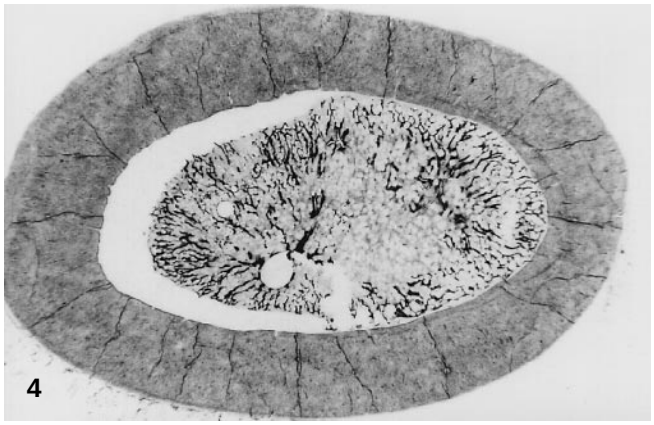
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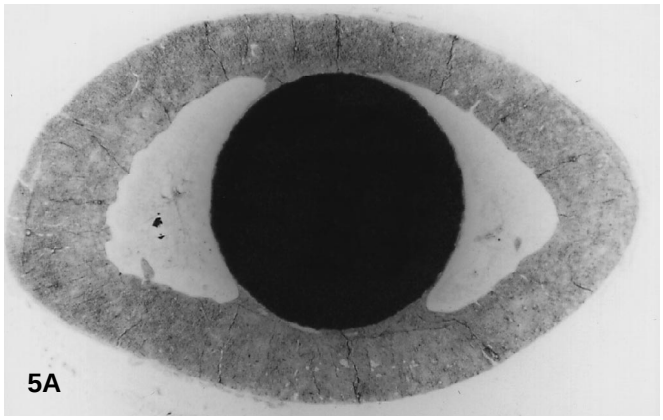
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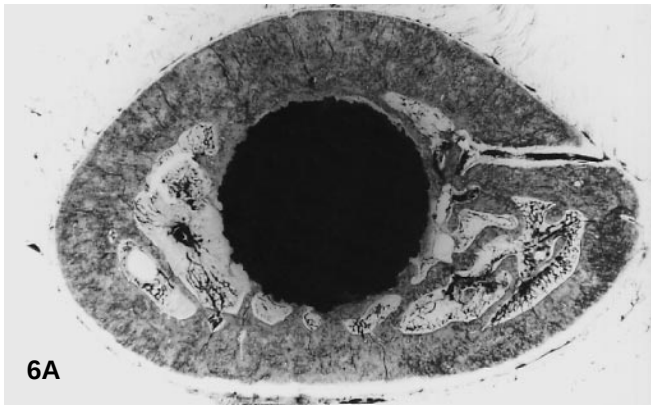
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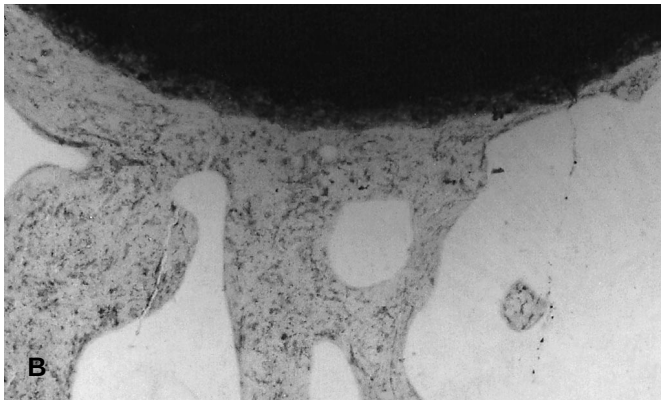
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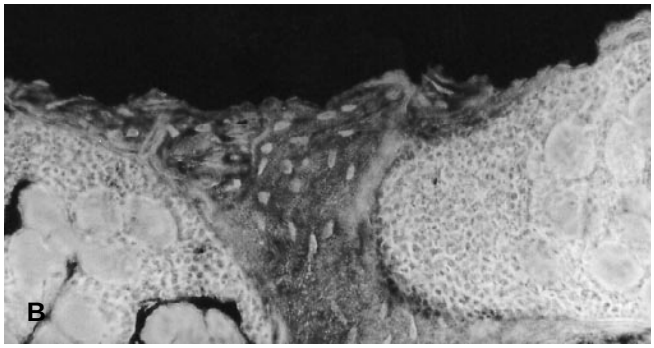
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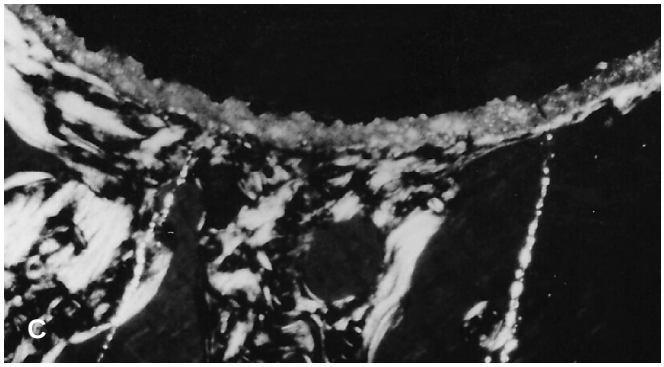
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Table 3 Morphometric evaluation of the bone integration percentage between HA- and Ti-coated implants in press-fit and non-interference-fit groups

	Femur		Mean of intersection I^+ ($I^+ \pm I^-$)	Integration (%)
Press-fit	Right HA-coated	$n = 5$	58 ± 4.18*	0.90
	Left Ti-coated	$n = 5$	34 ± 9.46*	0.53
Non-interference-fit	Right HA-coated	$n = 5$	0	0
	Left Ti-coated	$n = 5$	0	0

* $P < 0.001$, Student's *t*-test

the major axis of the bone. These sections were dehydrated in scalar ethanol solutions and then embedded in Technovit resin (Kulzer, Wehrheim, Germany), and thin sections were prepared with a cutting-grinding device (Exact Apparatebau, Norderstadt, Germany). Unstained sections 50 µ thick were observed under incident fluorescent light. Sections were then thinned to about 10 µ with the same cutting-grinding device, stained with haematoxylin eosin and observed under bright field transmitted light, polarized light and phase-contrast microscopy.

Morphometric evaluation of the implant surface in direct contact with bone was carried out with a reticulum characterized by radial disposition of lines passing through a central point; the angle between two adjoining radii was 5.625°. The centre of the reticulum was adjusted to the centre of the implant, and each intersection of radii with the interface was classified as I^+ if there was contact without soft-tissue interposition between implant surface and bone or I^- if not (Fig. 1). The degree of integration of each histological slide was expressed as the ratio of intersections type I^+ and the total number of intersections (integration percentage).

Only the press-fit and non-interference-fit groups at 90 days were evaluated for bone integration: the value of each femur was the mean of three slides perpendicular to the major axis of the femur, at distances of 6, 9 and 12 mm, respectively, from the intercondylar notch. The integration percentage between the right and left femur was compared in the same group and between groups with Student's *t*-test.

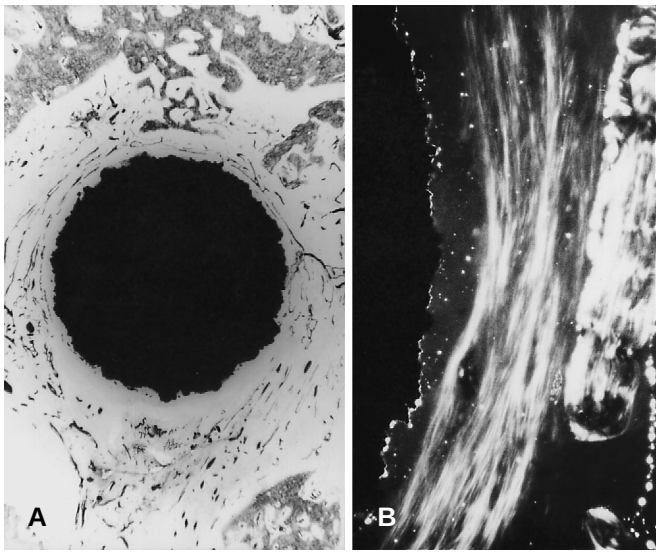

Fig. 7 A, B Over-reamed femur 90 days after implantation of Ti-coated rod and vessels injection. **A** No contact of endosteal bone with the implant surface is present; injected vessels show a circumferential arrangement around the implant (haematoxylin & eosin, × 40). **B** Collagen fibres of the connective capsule surrounding the implant show a circumferential arrangement (polarized light, × 250)

Results

Bone reaction to rod implantation at 15 days consisted of a circumferential apposition by the activated periosteum and new bone formation in the medullary space. Both types of response were present in the reamed and implanted femora and in controls (only reamed).

The periosteal reaction was characterized by hypertrophy of the periosteal vessels: these were deposited in a tangential manner on the bone surface and branched along a tortuous and radially oriented course toward the original cortex (Fig. 2). Primary bone formed around these vessels, characterized by wide osteocytic lacunae, plump osteocytes and a casual, three-dimensional arrangement of collagen fibres of the matrix.

The endosteum also reacted with the formation of primary, woven bone. Differentiation of osteoblasts occurred in the loose connective tissue which replaced the haematoma, but at this time interval the medullary vascular supply destroyed by the reaming procedure had not yet been reconstituted. Proliferation and activity of osteogenic cells in the medullary space was apparently not supported by vessel proliferation. From this point of view no relevant differences were observed between implanted and reamed control femora. Comparison of HA- and Ti-coated rods revealed a difference in the behaviour of osteogenic cells: in the former, rows of osteoblasts lined the implant surface and the HA coating served as substrate for matrix deposition, while in the latter, osteogenesis occurred in proximity to the surface, but direct apposition of matrix on the metal was never observed (Fig. 3).

At 90 days the periosteal primary bone has been completely remodelled, since labelled primary matrix was no longer evident and only scattered, forming osteons were labelled by tetracycline. The new periosteal layer was otherwise indistinguishable from the original cortex, and periosteal vessels had a normal aspect on the outer perimeter of the bone. Inside the medullary canal revascularization occurred in both reamed controls and implanted bones. A well developed net of injected vessels was present in the former, while the endosteal, reactive bone had been completely resorbed, and the only sign of the reaming procedure was a non vascularized scar in the centre of the medullary space (Fig. 4).

In the implanted femora revascularization of the medullary space was also observed; in these the reactive primary bone had been remodelled with formation of lamellar bone structured in radial bridges, connecting the endosteal surface with the rod.

In HA-coated implants a thin and continuous layer of lamellar bone encapsulated the outer perimeter of the rod

(Fig 5), while in titanium-coated implants the bridges formed extensions on the surface, but no continuous layer was present (Fig. 6).

The morphometric evaluation of the bone/implant contact is given in Table 3.

Over-reamed non-interference-fit femora, independently of the type of coating, were surrounded by a thick layer of fibrous tissue, with no bony connections with the endosteal surface. The collagen fibres of this tissue had a circumferential arrangement around the implant; vessels also showed a prevailing circumferential orientation (Fig. 7).

Discussion

Several factors are known to influence the bone growth on the surface of cementless prostheses: these include the bioactivity of coatings like HA and calcium phosphate [2, 3, 18], the porosity and roughness of the surface [20–22], the interface gap [17, 23] and the mechanical stability [13, 24]. The relative weight of each factor has not yet been ascertained, but there is a general consensus that absence of movement at the interface in the early period after prosthesis insertion is the fundamental condition for bone ingrowth or growth on the implant surface. This view was confirmed by the results of this study; in the experimental design employed implants were not subjected to weight-bearing conditions; this reduces considerably the stress and strain at the interface, which produce the micromotion to which clinical prostheses are subjected. However, interface strains were not completely abolished because of the different coefficient of elasticity of the femoral diaphysis and the intramedullary implant. Torsional and flexional deformations of the femur occur during walking: they are transmitted at the interface level of the intramedullary rod as strains in a tight press-fit implant or as movements in a no-fit situation such as the over-reamed and implanted femora of this experiment. Over-reaming by 0.3 mm produced inadequate fixation and sufficient micromotion to inhibit the bone integration of both HA- and Ti-coated implants.

Søballe et al. [17] have shown that the HA coating is capable of enhancing healing of a gap of about 1 mm at the bone-implant interface; however, in their experimental design a short implant lay in the spongious bone of the distal femural epiphysis and was placed perpendicular to the major axis of the bone, with lower strains than in a long intramedullary rod. Moreover, two spacers were fixed at each end of the implant to assure the presence of the spatial gap, but no movement at the interface. Therefore, comparison between the discrepancy of results of the Søballe experiment and that of the present study is not possible.

Since surface roughness, medullary location and the mechanical conditions were similar in the tight-fit and non-interference-fit groups, it can be assumed that stability is more relevant than the osteoconductive properties of the surface.

Several studies have demonstrated that the addition of a thin layer of plasma-sprayed HA on the surface of an implant has a favourable effect on the pattern and extent of bone formation in and around the implant [7, 8, 16, 17]; the present study confirms this, since there was a significantly higher percentage of surface covered by bone in HA-coated implants than in Ti-coated under the same mechanical conditions. However, the process of incorporation of an implant into bone is complex: it is influenced not only by the early reaction characterized by the production of primary, woven bone, but also by the subsequent bone remodelling. Two phases were observed in the process of implant integration.

First, an early phase, characterized by activation of endosteal osteoblasts or differentiation of new osteoblasts in the loose connective tissue, which replaces the haematoma produced by the reaming procedure. This type of response is independent of insertion of an implant in bone, since it was present with the same intensity and quality in the control animals, where reaming alone was performed. Histologically, it is characterized by the production of primary, woven bone; in our experimental model the development of this response is limited in time to the first 30 days following reaming and is homologous to the endosteal callus production of fractures [25]. In HA-coated implants osteoblasts adhere to the surface and lay down bone matrix on HA forming a true, chemical bond between the living tissue and the artificial substrate. In contrast, in Ti-coated implants osteoblasts were observed near the surface, but never arranged in apposition on the metal; therefore, only contiguity can exist, but no true bonding. The occurrence of matrix apposition by osteoblasts on a metallic surface, even titanium, has never been reported to the best of our knowledge and even direct contact between the bone mineral and metal surface was not demonstrated by ultrastructural studies of the interface [26]. Moreover, light microscopy and scanning electron microscopy (SEM) images of bone-titanium interface must be carefully interpreted because of the possibility of artefacts due to blurring and etching of the metal during the cutting procedures [27].

Second, a late phase, when the primary bone is remodelled to mature, lamellar bone. In controls (only reamed) all primary bone is resorbed, while for implanted femora it was structured in the form of a thin capsule surrounding the implant and of radial bridges connecting the latter to the endosteal surface. Since the remodelling process consists in resorption of the already formed bone and apposition of the new matrix in a lamellar pattern in the resorption lacunae, it follows that the structural organisation of the remodelled bone is conditioned by the quantity and distribution of the primary bone around the implant. A further factor controlling this remodelling is represented by mechanical forces: the distribution of the primary, reactive bone did not have a geometrical outline, while the lamellar bone was shaped in the form of radial bridges with extensions on the implant surface, which suggests a mechanical role of support and transmission of stresses. From this point of view, the type of organization of endosteal

lamellar bone, namely number, height and thickness of radial bridges, is fundamental to determine the mechanical condition of implant stability, more than the extent of the bony capsule around its perimeter.

This can explain the discrepancy which has been observed between the strength of the implant-bone interface measured by pull-out tests and the morphological parameters of ingrowth and growth of bone on the surface [8] and suggests that morphological evaluation of implant integration should also consider the radial connections between the implant and endosteal surface. These considerations are also supported by clinical observations of loosened, cementless stems where a layer of compact bone had grown on the porous surface, but there were no connections with the endosteum, and loosening occurred between the smooth capsule of bone on the stem and the endosteal surface [28]. The periosteal reaction is a well-known phenomenon subsequent to reaming of the medullary canal and has been extensively investigated in the context of intramedullary nailing for fractures [29–31]; it is related to hypertrophy of periosteal vessels in response to the damage to the intramedullary vascular supply and is characterized by a highly accelerated activity of osteogenic periosteal cells in a time interval of about 2 weeks after reaming. It is relevant in experiments concerning the endosteal fixation of cementless implants, because it produces first an increase of the outer perimeter of the bone and is followed later by an enlargement of the medullary canal, when medullary revascularization takes place. The consequence is a reduction or a loss of the endosteal fit of the implanted stem, whose stability rests more and more on the radial bridges between the implant and the endosteal surface. These aspects are clearly evident in experimental models involving small rodents like rabbits or rats where bone turnover and revascularization are very fast. Since the pattern of the vascular supply of long bones is the same as in larger animals or in humans, they must occur also in the latter, even if at a slower speed. They must be taken into consideration because they have a direct influence on the endosteal anchorage of the cementless implants, whether coated or uncoated.

The role of the vascular supply in osteogenesis is well-known [32]; paradoxically, the endosteal osteogenesis in the early phase after reaming was not accompanied by a demonstrable development of the medullary vascular net. A possible explanation could be that the blood supply was supported by vessels and capillaries too small to be injected by the perfusion technique. Medullary revascularization was evident in a later phase and was related to remodelling of the reactive endosteal bone. Formation of vascular spaces in the subendosteal portion of the cortex and in the mass of reactive bone inside the medullary canal is thus an inevitable consequence related to the biology of the bone [33, 34]; in this context it can influence the organization of the radial structures which support the mechanical stability of the implant.

In conclusion, this study confirms the view that HA coating has a favourable effect on bone apposition on the implant surface and gives evidence that this is due to the early adhesion of osteoblasts and direct deposition of bone matrix on the HA substrate. Moreover, it shows that the remodelling process is fundamental for implant fixation and stability in the long term; the latter is conditioned not only by the mechanical conditions and the physicochemical properties of the surface, but also by the biological response of the whole bone to changes in the vascularization pattern. Further studies are needed to define the relative role of these factors in the mechanism of implant integration in bone.

References

1. De Groot K, Geesink R, Klein CP, Serekian P (1987) Plasma sprayed coatings of hydroxyapatite. J Biomed Mater Res 21: 1375–1381
2. DeLange GL, DePutter C, DeWijs FL (1990) Histological and ultrastructural appearance of the hydroxyapatite-bone interface. J Biomed Mater Res 24: 829–845
3. Ducheyne P, Hench LL, Kagan A II, Martens M, Bursens A, Mulier JC (1980) Effect of hydroxyapatite impregnation on skeletal bonding of porous coated implants. J Biomed Mater Res 14: 225–237
4. Geesink RGT, de Groot K, Klein CPAT (1988) Bonding of bone to apatite-coated implants. J Bone Joint Surg [Br] 70: 17–22
5. Holmes RE, Bucholz RW, Mooney V (1987) Porous hydroxyapatite as a bone graft substitute in diaphyseal defects: a histometric study. J Orthop Res 5: 114–121
6. Moroni A, Caja VL, Sabato C, Egger EL, Gottsauner-Wolf F, Chao EYS (1994) Bone ingrowth analysis and interface evaluation of hydroxyapatite coated versus uncoated titanium porous bone implants. Mater Med 5: 411–416
7. Stephenson PK, Freeman MAR, Revell PA, Germain J, Tuke M, Pirie CJ (1991) The effect of hydroxyapatite coating on ingrowth of bone into cavities in an implant. J Arthroplasty 6: 51–58
8. Tisdel CL, Goldberg VM, Parr JA, Bensusan JS, Staikoff LS, Stevenson S (1994) The influence of a hydroxyapatite and tricalcium-phosphate coating on bone growth into titanium fibermetal implants. J Bone Joint Surg [Am] 76: 159–171
9. Turner TM, Sumner DR, Urban RM, Rivero DP, Galante JO (1986) A comparative study of porous coatings in a weight-bearing total hip-arthroplasty model. J Bone Joint Surg [Am] 68: 1396–1409
10. Hofmann AA, Bachus KN, Bloebaum RD (1993) Comparative study of human cancellous bone remodeling to titanium and hydroxyapatite coated implants. J Arthroplasty 8: 157–166
11. Bloebaum RD, Bachus KN, Rubman MH, Dorr LD (1993) Postmortem comparative analysis of titanium and hydroxyapatite porous-coated femoral implants retrieved from the same patient. J Arthroplasty 8: 203–211
12. Bloebaum RD, Bachus KN, Hamberg NG, Hofmann AA (1994) Mineral apposition rates of human cancellous bone at the interface of porous coated implants. J Biomed Res 28: 537–539
13. Søballe K, Hansen ES, Rasmussen HB, Jorgensen PH, Bunger C (1992) Tissue ingrowth into titanium and hydroxyapatite-coated implants during stable and unstable mechanical conditions. J Orthop Res 10: 285–299
14. Pazzaglia UE, Zatti G, Cattaneo S, Cherubino P (1983) Comparison of hollow and full stem implanted in the rabbit tibia. Preliminary results. Biomaterials 14: 883–886
15. Weinans H (1991) Mechanically induced bone adaptations around orthopaedic implants. Thesis: Cip-Gegevens, Koninklijke Bibliotheek, Den Haag

16. Søballe K, Hansen ES, Brockstedt-Rasmussen H, Pedersen CM, Bunger C (1990) Hydroxyapatite coating enhances fixation of porous coated implants: a comparison in dogs between press fit and non interference fit. Acta Orthop Scand 61:299–306
17. Søballe K, Haven ES, Brockstedt-Rasmessen H, Hjortdal VE, Juhl GI, Pedersen CM, Huid I, Bunger C (1991) Gap healing enhanced by hydroxyapatite coatings in dogs. Clin Orthop 272: 300–307
18. Brossa F, Cigada A, Chiesa R, Paracchini L, Consonni C (1993) Adhesion properties of plasma sprayed hydroxylapatite coatings for orthopaedic prostheses. Biomed Mater Eng 3: 127–136
19. Brossa F, Cigada A, Chiesa R, Paracchini L, Consonni C (1994) Post deposition treatment effects on hydroxyapatite vacuum plasma spray coatings. Mater Med 5:855–857
20. Bobyn JD, Pilliar RM, Cameron HV, Weatherly GC (1990) The optimum pore size for the fixation of porous surfaced metal implants of the ingrowth of bone. Clin Orthop 150:263–270
21. Cameron HV, Pilliar RM, McNab I (1976) The rate of bone ingrowth into porous metal. J Biomed Mater Res 10:295–302
22. Harris WH, Jasty M (1985) Bone ingrowth into porous coated acetabular replacements: the effect of pore size, apposition and dislocation. In: The hip, Proceedings of the 13th Open Scientific Meeting of the Hip Society. Mosby, St Louis, pp 214–234
23. Dalton JE, Cook SD, Thomas KA, Kay JF (1995) The effect of operative fit hydroxyapatite coating on the mechanical and biological response to porous implants. J Bone Joint Surg [Am] 77:97–110
24. Cameron HV, Pilliar RM, McNab I (1973) The effect of movement on the bonding of porous metal to bone. J Biomed Mater Res 7:301–311
25. Pazzaglia UE, Zatti G, Cherubino P, Frick W, Koch R (1991) Bone reaction to the implant of intramedullary pins of three different metals in rat femur. Mater Med 2:77–81
26. Sonnerby L, Thomsen P, Ericson LE (1992) Ultrastructure of the bone-titanium interface in rabbits. Mater Med 3:262–271
27. Pazzaglia UE, Bernini F, Zatti G, Di Nucci A (1994) Histology of metal-bone interface. Considerations about interpretation of plastic embedded slides. Biomaterials 15:273–279
28. Pazzaglia UE, Ricciardi C, Grassi FA, Chew Long EW, Scarponi R, Silvello L, Raimondi GB (1987) Risultati a medio termine delle protest non cementate (tipo Lord). Gior Ital Ortop Traum 13:501–506
29. Dawkwardt-Lillienstrom G (1969) Reaming of the medullary cavity and its effect on diaphyseal bone. Acta Orthop Scand [Suppl] 128
30. Dawkwardt-Lillienstrom G, Lorenzi GL, Olerud S (1970) Intracortical circulation after intramedullary reaming with reduction of pressure in the medullary cavity. A microangiographic study on the rabbit tibia. J Bone Joint Surg [Am] 52:1390–1394
31. Gothman L (1960) The arterial pattern of the rabbit's tibia after application of an intramedullary nail. A microangiographic study. Acta Chir Scand 120:211–219
32. Trueta J, Cavadias AX (1955) Vascular changes caused by the Kuntscher type of nailing. An experimental study in the rabbit. J Bone Joint Surg [Br] 37:492–505
33. Trueta J (1963) The role of the vessels in osteogenesis. J Bone Joint Surg [Br] 45:402–418
34. Rhinelander FW, Oxon MA, Nelson CL, Stewart RD, Stewart CL (1976) Experimental reaming of the proximal femur and acrylic cement implantation. Vascular and histologic effects. Clin Orthop 141:74–89