The design features of cemented femoral hip implants

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We undertook a review of the literature relating to the two basic stem designs in use in cemented hip replacement, namely loaded tapers or force-closed femoral stems, and the composite beam or shape-closed designs. The associated stem fixation theory as understood from *in vitro* studies and finite element modelling were examined with reference to the survivorship results for each of the concepts of fixation.

It is clear that both design principles are capable of producing successful long-term results, providing that their specific requirements of stem metallurgy, shape and surface finish, preparation of the bone and handling of the cement are observed.

Since the advent of cemented stem fixation in the early 1960s, surgical techniques and the design of implants have evolved dramatically. Some of these changes have resulted in improved survival while others have not.^{1,2} Mechanical studies *in vitro* and mathematical simulations with finite-element analysis are essential in evaluating the function of hip implants. However, predictions based entirely on these findings have not always been successful, illustrating the complexity of the interactions involved in the loosening of a cemented stem and the difficulty of reproducing the real-life situation.^{1,2}

We have reviewed the different surface characteristics of implants, the shapes of the stem and the various options in geometrical stem-broach sizing which have been adopted in an attempt to improve survival. In order to determine which of these combinations are satisfactory, experimental data have been related to clinical evidence of long-term success. The understanding of these concepts may help in the selection of potentially successful implants and of operative techniques.

Stem philosophies

The optimal shape of a stem should transmit torsional as well as axial load to the cement and to the bone without creating damaging peak stresses and without excessive micro-movement. The stem should remain mechanically stable in the long term despite being subjected to repetitive loading. Two methods have been adopted to achieve these goals: 'loaded-taper' or 'force-closed' fixation and

'composite-beam' or 'shaped-closed' fixation^{3,4} (Fig. 1).

In the loaded-taper model epitomised by the Exeter implant (Stryker, Mahwah, New Jersey), the stem is tapered in two (CPT, Zimmer, Warsaw, Indiana) or three planes (C-stem, DePuy International Ltd., Leeds, United Kingdom) and becomes lodged as a wedge in the cement mantle during axial loading,⁵⁻⁷ reducing peak stresses in the proximal and distal cement mantle.⁸ The stem is allowed to subside initially until "radial compressive forces are created in the adjacent cement and transferred to the bone as hoop stress".⁴ An air-filled distal centraliser is used to facilitate subsidence of the stem to a stable position without creating excessive stresses in the distal cement mantle.⁹

In the composite-beam concept, the stem needs to be rigidly bound to the cement since subsidence or impairment of the stem-cement interface may result in damage to the cement, with the generation of polymethylmethacrylate (PMMA) and/or metal debris and ultimately failure of the implant.¹⁰ Because these implants are not intended to subside, the presence of a void at the tip of the stem is considered to be detrimental since it weakens the cement mantle. Such voids can appear when air trapped by the implant in the recess meant to fix the distal centraliser undergoes thermal expansion during curing of the cement. For that reason, it is advised to always use a distal centraliser when such a hole in the stem is present or to plug that hole with the in-stem portion of the centraliser, a

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Fig. 1a

Fig. 1b

Fig. 1c

Radiographs showing stems of a) the loaded-taper type (CPT; Zimmer, Warsaw, Indiana) and b) the composite-beam type (Centralign, Zimmer) cemented undersized compared with the last broach, resulting in a thick cement mantle. Cementing a stem using a line-to-line technique gives a thin cement mantle and relies on c) the composite-beam fixation principle (Vectra; Biomet, Warsaw, Indiana).

plastic plug or cement which is allowed to cure before insertion of the stem.¹¹⁻¹⁴

Loaded-taper and composite-beam stems behave differently in terms of migration over time when studied by radiostereometric analysis (RSA). Loaded-taper stems in the first year of implantation show initial migration with reported mean subsidence ranging from 0.9 mm to 1.4 mm and retroversion between 0.4 mm and 0.5 mm.^{3,7,15-19} After the initial year, these stems tend to stabilise.^{3,15,17} Initial migration seems to be independent of the type of cement, its viscosity^{17,18} and the thickness of the cement mantle.¹⁷ Although tapered implants tend to stabilise only secondarily, they remain relatively stable over time.⁷ The cement mantle surrounding these stems does not migrate,^{3,16,19} or does so only slightly⁷ within the femur, which does not appear to compromise the long-term results. However, the degree of long-term migration which loaded-taper stems can tolerate is not known.⁷

Stems relying on the composite-beam principle have more initial stability especially in the longitudinal direction

with mean migration ranging between 0.1 mm and 0.5 mm during the first year.^{3,15,16,20,21} However, some tend to migrate also into retroversion, generally between 0.28 mm and 0.8 mm,^{15,20,21} but sometimes up to 1.0 mm and even 2.0 mm^{22,23} during the first year. In some instances migration at the cement-bone interface has also been seen.^{16,21,23} Both factors are worrying since excessive and continuous migration,^{23,24} may be considered to be predictive of failure.^{20,22}

Stem philosophy and surface finish

Polished stems are preferred with the loaded-taper design since they allow stepwise subsidence to a stable position,^{3,25} with the associated micromovement producing less metal and cement debris at the cement-stem interface.^{9,26,27} By contrast, in the composite-beam prostheses, it may be logical to optimise stability by roughening the surface to increase the cement-stem bonding.

Mechanical effect of the surface finish. From a mechanical perspective, a weak cement-stem bond with a polished stem

Table I. Details of studies showing the relation between the surface finish of cemented femoral hip implants with a similar stem geometry and clinical outcome

Authors	Implants*	Number of stems	Surface finish†	Ra (µm)‡	Mean follow-up (yrs)	Outcome of stem (%)§
Collis and Mohler ⁶⁷	Iowa	122	Polished	0.1	6.0	AL 0.0; RL 0.8
	Iowa	122	Matt + PMMA¶	2.1	5.3	AL 3.3; RL 1.6
	Iowa	36	Satin	0.8	11.3	AL 5.5; RL 5.5
Sporer et al ⁶⁸	Iowa	45	Matt + PMMA	2.1	8.2	AL 17.7; RL 4.4
Collis and Mohler ⁴⁸	Iowa	220	Satin	0.8	10.5	AL 3.5
	Iowa	343	Matt + PMMA	2.1	7.2	AL 3.1
	T-28	209	Polished	0.02 to 0.1	11.0	AL 1.9
	TR-28	227	Satin	0.8	10.5	AL 2.2
Meding et al ⁶⁹	T-28	378	Polished	< 0.1	12.4	AL 11.1; RL 0.0
	TR-28	171	Matt	1.6	10.8	AL 12.8; RL 3.0
Howie et al ⁷⁰	Exeter	20	Polished	0.02 to 0.04 ²⁷	> 9.0	AL 0.0; RL 0.0
	Exeter	20	Matt	1.03 to 1.83 ²⁷	> 9.0	AL 20.0; RL 5.0
Crawford et al ⁷¹	Exeter	56	Polished	0.01	9.0	AL 0.0; RL 1.8
	Exeter	23	Matt	1.2	9.5	AL 4.3; RL 13.0
Middleton et al ⁷²	Exeter	55	Polished	0.02 to 0.04 ²⁷	7.8	AL 0.0; RL 1.8
	Exeter	25	Matt	1.03 to 1.83 ²⁷	12.0	AL 24.0; RL 4.0
Dall et al ⁷³	Charnley first generation	264	Polished	0.02 to 0.03 ²⁷	8.8	AL 1.6; RL 1.6
	Charnley second generation	402	Satin-Matt	0.66 to 1.21 ²⁷	7.8	AL 6.7; RL 4.7
Kerboull et al ⁷⁴	Kerboull	165	Polished	0.03 to 0.05 ³⁷	9.0	AL 0.0
	Kerboull	141	Satin	0.6 ¹²¹	9.0	AL 1.5
Morscher et al ⁷⁵	MS-30	325**	Polished	-	8.8	AL 0.9; RL 24.1
	MS-30	586**	Satin-Matt	0.5 to 1.5	7.3	AL 0.7; RL 22.6
Della Valle et al ⁷⁶	Versys	138	Satin	0.5	5.4	AL 0.0; RL 2.2
	Versys	64	Matt	1.75 to 2.25	5.9	AL 10.9; RL 12.5

* Iowa, Zimmer, Warsaw, Indiana; T-28, Zimmer; TR-28, Zimmer; Exeter, Stryker, Mahwah, New Jersey; Charnley (first and second generation), DePuy International, Leeds, United Kingdom; Kerboull, unknown; MS-30, Zimmer; Versys, Zimmer

† according to Crowninshield et al⁶⁵

‡ Ra: average surface roughness (1 µm = 39.37 µinch)

§ AL, revision rate for aseptic loosening of the stem; RL, radiological loosening, osteolysis or radiolucent lines in unrevised hips

¶ PMMA, polymethylmethacrylate

** summary from four different series

has little effect on the distal, but increases proximal, cement strains.²⁸ When the cement-stem bond increases, compression stresses decrease, but higher tensile and shear stresses appear in the cement mantle and at the cement-bone interface.²⁹ Poorly-bound polished stems do not create tensile stresses and decrease shear stresses in the cement and the cement-bone interface.²⁹ In contrast to compression strains, which can be transmitted without a reliable cement-stem bond, transmission of tensile strains relies largely on a good bond.³⁰ Since this tensile bond may be unreliable over time, Crowninshield et al³⁰ have suggested that “it is unwise to design prostheses that rely heavily on the presence of a good (stem-cement) bond”. Moreover, because PMMA tolerates compressive loads well, but is more vulnerable to tensile stresses and shear forces,³¹ weakly-bound stems may load the cement in a less damaging way.

Failure of the implant due to accumulated mechanical damage to the cement-bone interface has been described both *in vitro*³² and *in vivo*.^{10,33,34} Because weakly-bound stems transfer less shear force and tensile stress to the cement-bone interface, that interface may be less damaged.²⁹ This explains why implants with a strong cement-stem bond may be more sensitive to the presence of incomplete and thin cement mantles with a poor cement-bone interface than polished stems.³⁴⁻³⁸

Wear and surface finish. Despite efforts to enhance the cement-stem bond by pre-coating or roughening of the stem, micromovement between the stem and cement seems to be inevitable. This is due to differences in elasticity between the stem, cement, cancellous and cortical bone and to the repetitive axial and torsional loads applied to the stem-cement-bone composite by the body-weight and by

Fig. 2

Diagram showing the influence of the stem-broach matching and cross-sectional shape of the stem on the appearance of the cement mantle (Capital Hip, 3M Health Care Ltd, Loughborough, United Kingdom; Centralign, Zimmer, Warsaw, Indiana; Kerboull CMK II, Smith & Nephew, Memphis, Tennessee; Kerboull CMK III, Vecteur Orthopédique, Marne la Vallée, France; Vectra, Biomet, Warsaw, Indiana; Exeter, Stryker, Mahwah, New Jersey; Charnley, DePuy International Ltd, Leeds, United Kingdom; Kerboull CK I, Stryker; Kerboull CK II, Stryker; Müller, Zimmer).

the action of the muscles on the proximal femur.^{39,40} Mechanical testing *in vitro*,⁴¹⁻⁴⁴ as well as RSA studies *in vitro* and *in vivo*,^{35,45,46} have shown that perfect stability of the stem is improbable. When debonding finally occurs at the pre-coated or roughened cement-stem interface, there will be damage to the cement and large quantities of PMMA and/or metal particles will be generated^{25,47} causing osteolysis, loss of bony support and loosening of the implant.^{10,26,36,48-53} This seems less critical for polished stems because, compared with non-polished implants, gap formation at the cement-stem interface²⁵ and migration of particles along that interface are reduced,⁵⁴ and micro-movement between the implant and cement produces less debris.^{25,27,47,55} Moreover, whereas unpolished stems become polished at the surface with release of debris, polished stems mostly show pitting, with retention of debris on their surface.²⁷

Wear at the cement-stem interface will be even greater if unpolished stems, made of materials with less wear-resistance such as titanium alloy, loosen.^{53,56-58} This may

account for the poor performance of unpolished cemented titanium stems^{33,51,56,59-63} while polished or smooth cemented stems of this material have survived well.^{37,64-66}

It can be concluded that 'rougher' stems need a thick, continuous cement mantle of good quality with a strong cement-bone interface and should be made of wear-resistant materials, whereas polished stems may be more tolerant to suboptimal cementing and manufactured from less wear-resistant materials. This may also explain why the same design of stem but with a smoother surface finish performs better than its rougher equivalent, even if they are of the composite-beam design (Table I).^{48,67-76}

Geometry of the stem. Several features of the shape of the stem influence the *in vivo* behaviour of femoral components, including the overall shape (straight or anatomical), the cross-section (oval or square), the presence of a collar, the shape of the tip of the stem, the length of the stem and whether the edges are rounded to a greater or lesser degree. **Straight and anatomical stems.** In contrast to symmetrical stems¹⁴ such as the Charnley (DePuy International Limited,

Leeds, United Kingdom), Exeter (Stryker), Müller (Zimmer, Warsaw, Indiana), Versys (Zimmer) and Spectron (Smith & Nephew, Memphis, Tennessee), anatomically-shaped components like the Lubinus SP2 (Waldemar Link GmbH, Hamburg, Germany), ABG (Stryker), Olympia (Biomet, Warsaw, Indiana), Aura II (Biomet), SHP (Biomet) and APR II (Zimmer) are designed to fit the sagittal intra-medullary anatomy.⁷⁷ This allows better centralisation of the stem and more even thickness of the cement mantle.⁷⁸ Compared with symmetrical stems, anatomical stems generate different strains within the cement mantle because of their specific shape.⁷⁹ They are of the shape-closed or composite-beam type since their shape limits the subsidence required to achieve a stable position. However, it remains questionable if these characteristics are key advantages since both types of stem have performed equally well in the long term.^{9,46,80-83} Nevertheless, an anatomical stem, which can be inserted more anteriorly in the shaft without creating an area of anterior proximal cortical contact and posterior distal point contact, could be an advantage, especially for the less experienced surgeon using an anterior approach.

The cross-sectional shape of the stem. The cross-sectional shape influences the distribution of cement within the femoral canal, the rotational stability of the implant⁸⁴ and the stress distribution within the cement mantle.³⁰ Broaches and stems with an oval cross-section as found in the Kerboul CMK II (Smith & Nephew) and III (Vecteur Orthopédique, Marne la Vallée, France), Vectra (Biomet) and Centralign (Zimmer) have a better fit within the medullary canal and can occupy more of the cavity, leaving less room for cement and cancellous bone. By contrast, the broaches and stems with a more rectangular cross-section such as the Exeter (Stryker), CPT (Zimmer), CPS-plus (Endoplus, Swindon, United Kingdom) and the Kerboul CK I (Stryker), are limited in size by their contact against the inner cortex of the oval cross-section of the medullary canal. This may result in additional space for pressurisation of cement into the remaining cancellous bone beyond the reach of the broaches. However, if cement is not or cannot be fully pressurised into that layer, mechanically weak cancellous bone will be interposed between the cement and cortical bone. This can be avoided by removing this cancellous bone with a curette before cementing the stem, taking care to leave a minimal amount of well-fixed cancellous bone attached to the cortex to allow proper interdigitation of the cement (Fig. 2). Since modern cementing techniques allow pressurisation of cement into cancellous bone over a distance of 3 mm,⁸⁵ that amount of remaining cancellous bone should be adequate.

Stems with a square cross-section offer more rotational stability than oval stems. However, sharp edges create peak stresses in the cement, which could lead to microfractures. Mann and Kim⁸⁴ calculated, based on a finite-element model, that optimal rotational stability with acceptable peak stresses in the cement was obtained when the corners had a fillet radius of 2 mm. In clinical practice, the original

polished Charnley-Kerboul stem with a quadrangular cross-section performed better than a later matt version with an oval cross-section.^{37,86} However, it is unclear whether that underperformance could be attributed to the oval cross-section and/or the matt surface finish.

The addition of proximal anteroposterior cobra-shaped dorsal flanges to the stem has been advocated in order to decrease stress shielding, to enhance stability of the stem and to increase the interlock between the stem and the cement.⁸⁷ However, dorsal flanges caused higher cement-bone micromovement *in vitro*⁸⁷ and are associated with more cement-bone radiolucencies *in vivo*.⁶⁰ In clinical practice, a dorsal flange reduced subsidence of the stem and the incidence of fractures of the distal cement.⁸⁸ However, the survival rates of the flanged grit-blasted cobra-shaped Charnley stem at 15 and 25 years were less satisfactory than those of the non-flanged polished version.⁶⁰ It is unclear if this was attributable to the flange, the rougher stem surface or to both.

In order to improve the rotational stability of polished tapered stems, the CPS-plus stem (Endoplus) was designed to fill the canal to a greater extent and had a broader shoulder and a more rectangular cross-section compared with the original Exeter stem (Stryker). Two years after implantation RSA confirmed a similar pattern of subsidence but with improved rotational stability and decreased valgus migration compared with the original Exeter design.⁸⁹ This is expected to improve the long-term outcome but remains to be proven by clinical follow-up studies.

Geometry of the proximal stem. Experimental^{79,90-92} and finite-element analysis studies^{29,90,92,93} both found high focal strains in the cement mantle at the level of the medial femoral neck and/or near the tip of the implants. These regions are vulnerable to damage to the cement during initial loading.⁹⁴ When cracks in the cement extend from the region of the medial femoral neck and from the tip of the stem over the complete length of the stem, failure of the implant is imminent.⁹⁴ Therefore, it is important to reduce cement strains in these regions, and to obtain a cement mantle of good quality removing mechanically-weak cancellous bone between the cement and the cortex, especially in the region of the medial femoral neck. This last point has been supported by mechanical testing,^{95,96} finite-element analysis⁹⁷ and clinical data.⁹⁸⁻¹⁰²

Theoretically, the collar of the cemented femoral component has two functions. First, it has the potential to promote direct transfer of load from the implant to the medial cement mantle and/or the bone of the medial femoral neck, at least when a close contact is achieved initially and maintained over time. Moreover, direct collar-bone contact can unload the vulnerable proximal cement mantle.^{91,103} The presence of a collar can also reduce tensile stresses in the stem¹⁰⁴ and reduce overall migration.^{21,105} However, a collar has a negative effect on the final rate of migration,¹⁰⁵ preventing the stem from 'settling' during cyclic loading, and does not avoid micromovement of the

Table II. Stem-broach mismatch of different femoral stems according to a manufacturer's survey performed by us in 2002

Stems > 1 mm undersized compared with broach		Stems ≤ 1 mm undersized compared with broach	
Brand*	Stem	Brand*	Stem
Zimmer	CPT	Zimmer	Metabloc
Zimmer	Versys	Zimmer	Müller
Zimmer	Harris	Zimmer	Exofit
Zimmer	MS 30	Zimmer	Weber
DePuy	Charnley, Elite/Elite plus	DePuy	Modulor
DePuy	C-stem	Céraver	Ostéal
Stryker	Exeter	Stryker	Kerboull MK III
Stryker	Contemporary hip	Stryker	Legend V40
Endoplus	CPS-Plus	Stryker	ABG cemented
Biomet	Mallory Head interlock	Biomet	Vectra III
Biomet	Stanmore	Biomet	Aura II
Biomet	Answer, Alliance	SEM	SEM III
Wright Medical	Helianthe	Wright Medical	PTC
Smith & Nephew	Spectron	Waldemar Link	Lubinus
Smith & Nephew	Synergy	Smith & Nephew	CMK 21

* Zimmer, Warsaw, Indiana; DePuy International Ltd, Leeds, United Kingdom; Stryker, Mahwah, New Jersey; Endoplus, Swindon, United Kingdom; Biomet, Warsaw, Indiana; Wright Medical, Arlington, Tennessee; Smith & Nephew, Memphis, Tennessee; Céraver, Roissy, France; SEM, Montrouge, France; Waldemar Link GmbH, Hamburg, Germany

stem¹⁰⁵ or the production of wear debris from the cement-stem interface.²⁷ Neither does a collar prevent early resorption of the medial femoral neck,¹⁰⁶⁻¹¹⁰ which could be due to debris generated by attrition of the collar against bone and cement,¹⁰⁸ and could jeopardise the loading function of the medial femoral neck. A collar is also counter-productive in loaded-taper stems since they need to subside within the cement mantle to reach a stable final position.¹⁰⁸

The second function of the collar is to control insertion, especially when the stem is undersized compared with the broach, so that the final implant is inserted to exactly the same level as the broach. For implants which are not undersized, this is not crucial since the stem will automatically be directed to the broach position by contact with the bone.

A collar should only be considered in composite-beam stems which are undersized compared with the broach. However, even then, survival of the stem has not been improved as demonstrated in series comparing the same geometry with and without a collar.^{111,112}

Stem-broach mismatch. From a biomechanical^{92,97,113,114} and clinical^{99,115,116} point of view it has been recommended that a cement mantle which is subjected to high stresses should be between 2 mm and 5 mm thick, especially in the proximomedial part of the implant and around the tip of the distal stem. However, cement mantles thicker than 5 mm to 10 mm could increase micromovement and could be detrimental.^{99,113} Retrieval studies have reported more cracks in areas of thin cement.¹¹⁷ Mechanical and finite-element studies of the propagation of fatigue cracks in the cement showed that the rate of growth of the crack was independent of the thickness of the cement mantle.^{114,118} However, cracks in thin cement reached full-thickness in fewer loading cycles.¹¹⁸ After loading of cemented stems, regions of thin cement (< 2 mm) presented fewer cement

cracks but more full-thickness cracks than the other regions.¹¹⁴ Finally, from a biological point of view full-thickness cracks together with defects in the cement constitute a possible pathway for migration of particles from the cement-stem interface to the bone. This could be a source of particle-induced osteolysis even in well-fixed implants.^{26,38,119,120}

In order to favour a 'thick and flawless' cement mantle, some systems use stems which are undersized compared with the corresponding broach. However, in addition to these undersized stems most manufacturers also market stems which have the same size as the broach system and which are cemented 'line-to-line' (Table II).

Recently, the old debate concerning the appropriateness of undersizing the stem has been raised again by Langlais et al^{37,121} as the 'French paradox'. These authors, among others,^{64,86} have presented and reviewed excellent long-term results obtained with different polished and rectangular canal-filling stems cemented line-to-line after using the largest possible broach. These stems aim at a direct load transfer to bone by close cortical contact. As such, they are not meant to subside within the cement mantle and can be considered as 'shape-closed'. Comparing the results at ten years of the Freeman hip replacement, cemented line-to-line or undersized, Skinner et al¹²² concluded that a line-to-line technique "is not worse and may produce better long-term results than current teaching suggests". Müller¹²³ reported similar findings, and 10 to 15 years after introducing his straight stem he noted that: "The closer the contact between the stem and the bone, the better were the results". Therefore, despite a potentially suboptimal cement mantle, some stems inserted line-to-line with the largest broach have performed very well clinically (Table III).^{66,74,86,100,122,124-128}

Canal-filling stems offer many theoretical advantages. First, removal of a maximal amount of mechanically-weak

Table III. Details of studies reporting results of femoral-canal-filling stems inserted with a minimal cement mantle

Authors	Implants*	Number of stems	Cross-section	Surface finish†	Ra (in μm)‡	Material	Mean follow-up (yrs)	Outcome of stem (%)§
Delaunay et al ¹²⁴	Kerboull	215	Rectangular and oval	Polished and satin	0.03 and 0.6 ¹²¹	Stainless-steel	14.2	AL 2.8; RL 5.6
Kerboull et al ⁸⁶	Kerboull MKI & MKIII	166	Rectangular	Polished	0.03 ¹²¹	Stainless-steel	14.5	AL 0.6; RL 4.9¶
	Kerboull CMKII	51	Oval	Satin	0.6 ¹²¹	Stainless-steel	14.5	AL 10.0; RL 4.9¶
	Kerboull CMKIII	70	Oval	Satin	0.6 ¹²¹	Stainless-steel	14.5	AL 3.9; RL 4.9¶
Kerboull et al ⁷⁴	Kerboull	141	Oval	Satin	0.6 ¹²¹	Stainless-steel	9.0	AL 1.5
Arama et al ¹⁰⁰	Kerboull	481	Rectangular	Polished	0.03 ¹²¹	Stainless-steel	250 hips > 5.0	AL 0.0; RL 2.8
Delaunay et al ¹²⁴	Céraver Ostéal	165	Rectangular	Smooth	0.2**	Titanium	8.0	AL 2.4; RL 11.2††
Rousseau et al ¹²⁵	Céraver Ostéal	104	Rectangular	Smooth	0.2**	Titanium	11.0	AL 1.0; RL 2.9
Hamadouche et al ¹²⁶	Céraver Ostéal	89	Rectangular	Smooth	0.2**	Titanium	19.7	AL 12.7; RL 63.4††
Le Mouel et al ¹²⁷	Céraver Ostéal	156	Rectangular	Smooth	0.2**	Titanium	7.4	AL 0.0; RL 1.7
	Céraver Ostéal	61	Rectangular	Smooth	0.2**	Titanium	5.3	AL 0.0
Osorovitz and Goutallier ¹²⁸	Céraver Ostéal	124	Rectangular	Smooth	0.2**	Titanium	5.4	AL 0.0; RL 2.5
Nizard et al ⁶⁶	Céraver Ostéal	187	Rectangular	Smooth	0.2**	Titanium	10.9	AL 0.7; RL 3.0
Skinner et al ¹²²	Freeman	97	Mixed‡‡	Mixed§§		Cobalt-Chrome	10.0	AL 1.2; RL 7.3

* Kerboull, unknown; Kerboull MKI & MKIII, Stryker, Mahwah, New Jersey; Kerboull CMK II, Smith & Nephew, Memphis, Tennessee; Kerboull CMK III, Vecteur Orthopédique, Marne la Vallée, France; Céraver Ostéal, Céraver, Roissy, France; Freeman, Corin Medical Ltd, Cirencester, United Kingdom and Finsbury Instruments Ltd., Leatherhead, United Kingdom

† according to Crowninshield et al⁵⁵

‡ Ra: average surface roughness (1 μm = 39.37 μinch)

§ AL, revision rate for aseptic loosening of the stem; RL, radiological loosening, osteolysis or radiolucent lines in unrevised hips

¶ data mentioned are for all stem types as a group

** values provided by the manufacturer

†† including radiolucent lines < 1 mm or non-evolving radiolucent lines

‡‡ proximally, rectangular and neck preserving; distally round

§§ proximally, textured and sand-blasted; distally, polished

cancellous bone favours direct load transfer to the cortex. This occurs either through direct point contacts between the implant and cortical bone or through a thin cement layer between the implant and cortex without interposition of weak cancellous bone. This concept may improve the stability of the implant.^{29,30} Secondly, points of contact between implant and cortex could facilitate stem insertion by controlling stem alignment and insertion depth and by stabilising the implant during cement curing. Finally, insertion of a stem with similar dimensions as the largest possible broach creates high intramedullary cement pressures,¹²⁹ which could favour interdigitation of the cement into the remaining cancellous bone and cement penetration up to the cortex. Since larger quantities of cancellous bone remain in the proximal part of the femur after broaching, the effect is more marked in that region.¹²² This could improve the quality of the cement mantle, especially if a suboptimal cementing technique is used without adequate pressurisation. Therefore it is possible that cemented line-to-line implants could be more 'user-friendly' for less experienced surgeons.

A recent study⁸⁵ using CT and polymeric Charnley-Kerboull replicas showed that these stems, when cemented line-to-line with the largest broach, created a cement mantle which averaged over 3 mm in thickness. Because of pressurisation of the cement into cancellous bone, defects (< 1 mm of cement thickness) were found in only 6% of the interface, but areas with a cement thickness < 2 mm were found in 26%. These were noted mostly in the distal two-thirds and at the corners of the stem. Because areas of thin cement appeared to be supported mainly by cortical bone, they might be less detrimental than initially believed.

Summary and conclusions

Cemented femoral implants have been developed to function either as loaded-tapers or composite-beams. Stems of the loaded-taper type should be polished to favour stepwise subsidence to a stable position. They are very sensitive to a rough surface finish and are incompatible with the use of a collar as a positioning device, an anatomical shape or canal-filling design of the stem, since these features prevent subsidence within the cement mantle.

A stem relying on the composite-beam principle can be either straight or anatomical. Both have proved to be equally successful. Composite beams can be achieved with the interposition of a thick or a thin layer of cement, depending on whether the implant is undersized compared with the broach or not. A canal-filling stem is cemented line-to-line with the size of the last broach used and stem-cortex contact points as well as areas of thin cement supported by cortical bone help to stabilise the implant. In clinical practice, this user-friendly concept has been proved to work well.

In the composite-beam design the use of a rough surface finish to increase the stability of the stem seems to be logical but it can generate detrimental cement and metal debris when the inevitable micromovement of the stem appears. This is troublesome especially when a canal-filling stem is used, since particles will be allowed to migrate to the cement-bone interface through areas of cortical point contact and to create osteolysis. Additionally, a rough surface finish increases tensile and shear stresses in the cement mantle and at the cement-bone interface. This makes rougher stems less forgiving with suboptimal cementing conditions. Finally, a rough surface finish is not compatible with less wear-resistant materials such as titanium. In clinical practice, polished stems with a similar design were found to be more reliable in the long term compared with rougher stems and a smooth or polished surface finish should be favoured even for implants relying on the composite-beam principle.

Excessive and continuous migration of the implant is detrimental for both loaded-taper and composite-beam stems. However, it seems that implants of the loaded-taper type tolerate initial migration better and present a more stable cement mantle at the cement-bone interface. Composite-beam stems tend to be globally more stable, especially in the first years. However, in some cases, rotational instability may appear and excessive load transfer to the cement mantle may result in migration of the cement at the cement-bone interface. Both these phenomena are known to be predictors of poor long-term results.

Although *in vivo* both concepts of stem fixation have proved to be effective, they cannot work together. It is important to understand on which principle a particular stem relies.

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