



Review article

EVOLUTION OF THE INTERNAL FIXATION OF LONG BONE FRACTURES

THE SCIENTIFIC BASIS OF BIOLOGICAL INTERNAL FIXATION: CHOOSING A NEW BALANCE BETWEEN STABILITY AND BIOLOGY

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The internal fixation of fractures has evolved in recent decades with a change of emphasis from mechanical to biological priorities. More flexible fixation should encourage the formation of callus while less precise, indirect reduction will reduce operative trauma. This approach is described as 'biological internal fixation'. It involves the use of locked internal fixators which have minimal implant-to-bone contact, long-span bridging and fewer screws for fixation. Formerly, internal fixation with a plate aimed at absolute stability to avoid micromovement which could result in loosening of the implant and a delay in healing. The new technique of internal fixation, however, seems to tolerate and even require some degree of mobility of the interface of the fracture. The contact between implant and bone is kept stable using screws which function like locked threaded bolts.

The concept of biological internal fixation is still developing. This review focuses on the biomechanics of stability, the biology of the blood supply and the impact on clinical use.

The surgical treatment of fractures underwent an important change around the middle of the last century. Stable internal fixation allowed fractures to unite while maintaining the function of the joints and soft tissues. The positive aspects of internal fixation using compression techniques were restoration of the precise anatomy and early function. Intra-articular fractures could be reduced and stabilised with smooth and congruent joint surfaces which improved the prospect of avoiding post-traumatic arthritis. Precise reconstruction and absolute stability of fixation were considered to be essential preconditions for success.^{1,2} The evolution of conventional plate fixation and its present status have recently been summarised.³

Conventional stable internal fixation with precise reduction usually requires a fairly extensive surgical approach to

the bone. This contributes to increasing the necrosis which has been initially produced by the injury, consequently enhancing the risk of delayed healing, infection and possibly refracture. A demanding degree of skill and expertise is required at operation to minimise the biological complications following extended traumatic and iatrogenic necrosis.⁴

Recent developments aim to produce minimal biological damage with flexible fixation.⁵⁻¹⁰ 'Bio-logical',¹¹ internal fixation avoids the need for precise reduction, especially of the intermediate fragments, and takes advantage of indirect reduction.^{12,13} This principle applies equally to locked nailing,¹⁴⁻¹⁶ bridge plating,¹⁷⁻¹⁹ and internal fixator-like devices.²⁰⁻²⁸ Indirect reduction aims only to align the fragments. It avoids exposure of the bone thus reducing the surgical trauma. Flexible fixation^{18,29,30} is advocated to induce formation of callus^{31,32} and is achieved by using wide bridging¹⁹ of the area of the fracture. Pure splinting without compression results in flexible fixation. The avoidance of biological damage produced by overly precise reduction, the application of too many implants and too extensive implant-to-bone contact should reduce the risk of biological complications and improve healing. The aim is to produce the best biological conditions for healing rather than absolute stability of fixation and this approach has been shown to give early solid union.³³ Restoration of function is the principal object of both non-operative³⁴ and operative¹ treatment.

Biological internal fixation (Fig. 1) does not compromise the restoration of early and complete function of the bone, limb and patients, but recognition of the optimum requirements for bone healing now takes precedence, with mechanical stabilisation being less rigid while still allowing painless function and reliable healing. The aim is to reduce the infrequent but possibly severe complications⁴ such as sequestration and infection which may be produced by bone necrosis, with less emphasis on avoidance of delayed or nonunion, which is more easily managed. Knowledge of the scientific background to this more flexible biological approach will allow selection of the proper balance between mechanical and biological priorities according to the individual situation. For the basic information on bone biology and its clinical application reference should be made to a recent review.³⁵ The vascular aspects of bone healing are addressed in earlier^{36,37} and more recent papers.³⁸⁻⁴²

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Fig. 1a

Fig. 1b

Fig. 1c

Fig. 1d

Fig. 1e

Radiographs showing the evolution of fracture treatment. Various methods give two different radiological types of healing (indirect and direct) depending on the biomechanical environment which is produced by the method of stabilisation (relative stability, absolute stability). Figure 1a – Spontaneous healing in a mountain goat. There was no treatment and solid indirect union with severe malalignment resulted. Figure 1b – Conservative treatment with solid indirect healing in an acceptable position. This is a type of treatment which is still widely used. Figure 1c – Absolute stability produced by lag screws with direct healing in precisely restored anatomy. Figure 1d – Extensive compression plate fixation of a distal tibial fracture. There is healing without external formation of callus. The required approach for the combined use of cerclage wire, lag screws and plate screws is associated with the risk of producing extensive bone necrosis. Figure 1e – ‘Biological internal fixation’ of bilateral femoral fractures using an interlocked nail and bridging plate. The intermediate fragments were not touched. The patient was freely weight-bearing after eight weeks. Postoperative (mediolateral) and one year (AP) radiographs.

The results using biological internal fixation are impressive but it is only of value in the presence of living bone. The repair of dead bone requires long-term stability to allow for creeping substitution and internal remodelling, and therefore the combination of necrosis and instability may be deleterious.

The scientific background of biological internal fixation will now be discussed under the following headings.

Aspects of stability. The reason why instability may be detrimental in some instances and beneficial in others is assessed as well as the critical parameter of instability, namely strain.

Potential benefits of avoiding necrosis. Based on new insights, early temporary porosity, sequestration, infection and refracture are reviewed.

Consequences for surgical technology. Changes in the principles, procedures, instruments and implants are considered. The potential benefit of the new technology for the

surgical treatment of fractures in porous bone is given special attention.

Differential indications for conventional versus biological internal fixation. The conditions under which one method is superior to the other are discussed.

Requirements for further development. Scientific input is required to improve the indication, the principles and techniques of use.

The more general scientific aspects of bone,^{35,43-45} the current status of internal fixation,⁴⁶ special consideration of bone loading, blood supply and also intracortical fluid flow have already been discussed in special overviews.^{36,38,40-42,47-50}

Aspects of stability

The term stability is applied here according to its use in clinical practice, namely to define the degree of load-

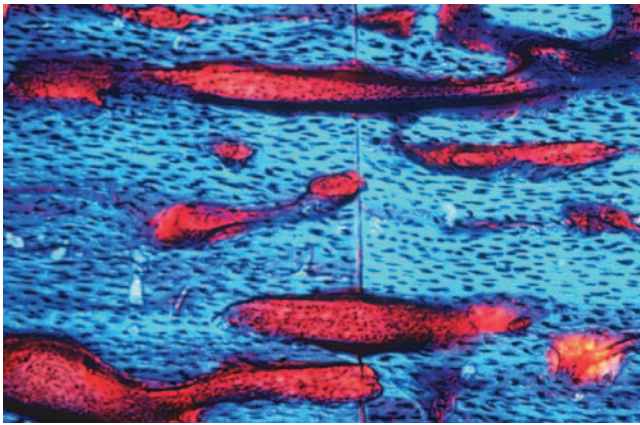


Fig. 2

Direct healing and induction of remodelling. The osteotomy planes are closely adapted and compressed and there is no displacement at the site of the osteotomy. Osteotomy of the tibia of a sheep with compression stabilisation. The shape of the osteons does not change when crossing the fracture or osteotomy. This strongly suggests that the mere presence of a fracture without dynamic microdisplacement neither influences nor induces remodelling.

dependent displacement of the fracture surfaces. When the interfaces of a fracture are compressed, no displacement may be observed, indicating absolute stability of fixation. The surfaces of fractures which have been splinted by implants without application of compression undergo relative displacement. This is proportional to the load applied and inversely proportional to the stiffness of the device used.⁴⁸ If rigid internal fixation depends on the avoidance of instability, why then can biological internal fixation take advantage of elastic flexible fixation?

Stability of fixation and type of healing of the fracture. Danis⁵¹ observed that after compression fixation the fracture healed without radiologically visible callus, which indicated that there was a close link between stability and the type of healing. However, the bone regeneration achieved using flexible fixation as described by Ilizarov⁵² is outstanding.

Indirect versus direct healing. Indirect healing consists of the sequential steps of tissue differentiation, resorption of the surfaces of the fracture and uniting of the fracture fragments by callus. Finally, the fracture undergoes long-lasting internal remodelling.⁵³ This is the pattern of healing without stabilisation, with stabilisation by an external or internal fixator⁵⁴ and with flexible internal fixation.⁵⁵

Direct healing follows stable fixation and compression; the bone heals without apparent callus (*soudure autogène*,⁵¹ primary healing⁵⁶). It skips the intermediate steps of tissue differentiation and resorption of the bone surface and progresses directly, although not necessarily more quickly, to the final internal remodelling of the Haversian system.



Fig. 3a



Fig. 3b



Fig. 3c

Radiographs showing unstable plate fixation resulting in nonunion. Do such observations challenge biological internal fixation using flexible fixation? What are the means of avoiding this evolution when practising biological internal fixation? The nonunion in this case is clearly the result of the implants not providing stability because of an obstructing lag screw, together with extensive traumatic and/or iatrogenic necrosis. Figure 3a – Inefficient nearby placement of the screw in both main fragments. Figure 3b – After eight months there is nonunion with malalignment. The lag screw was placed between the fragments and produced unstable obstruction. Figure 3c – Repair of the nonunion by plate bridging. There is prompt solid union within three months.

Because the radiographic picture does not disclose what happens within a closely adapted and compressed fracture gap, the monitoring of healing is indirectly based on the absence of adverse clinical and radiological symptoms. The appearance of callus and of bone resorption at the interfaces after 'stable' internal fixation was understood to indicate that the stability achieved did not match that intended.

With absolute stability of fixation the Haversian osteons cross the plane of contact of the fracture without obvious change in shape or direction (Fig. 2). The same holds true for stably fixed serial microfractures.⁵⁷ The question arises as to how a stably fixed fracture proceeds to remodelling. Is local necrosis the stimulus for this final phase of repair? Direct healing yields fascinating histological observation. It is rarely a goal in itself but rather a product of maintained absolute stability.

Deleterious effects of instability after rigid internal fixation. Biomechanical experiments have shown that in cortical bone in sheep a displacement of even a few micrometres at an interface between implant and bone, or bone and bone, induces resorption of the bone surface.⁵⁸ In contrast to the generally accepted view that resorption at interfaces to implants is the consequence of the breakdown of bone under excessive load, in these experiments both the displacement and the load were minimal. Figure 3 shows that after rigid internal fixation such resorption may endanger the result of the treatment because loosened implants may produce unstable obstruction, a type of instability which is difficult to overcome without providing improved fixation^{59,60} (Figs 3 and 4).



Fig. 4a

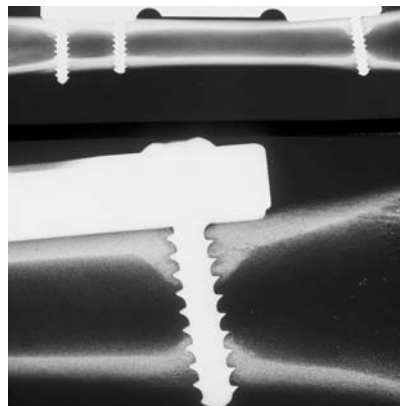


Fig. 4b

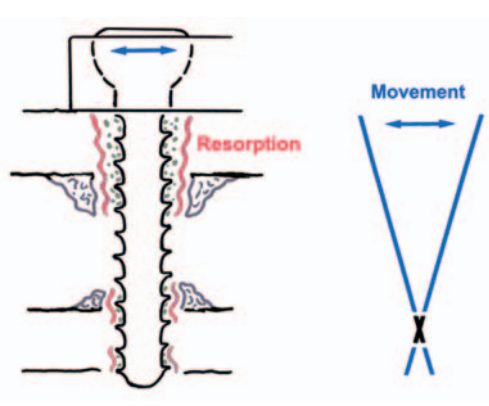


Fig. 4c

Clinical and experimental observations regarding unstable interfaces. Figure 4a – Radiograph showing a bone screw splinting two bones which move in relation to each other (syndesmosis screw at the left). This screw shows marked resorption while that compressing a stabilised fracture gap at the medial malleolus does not induce resorption in spite of the fact that the forces acting on the thread in the latter case are higher, but do not produce micromovement. Figure 4b – An instrumented plate (above) allows monitoring of the force transmission to the single screw on the right bridging a segment of a sheep tibia. The two screws on the left lock that end of the plate tightly to the bone. When the sheep stands on this leg, the tibia shortens by a few micrometres. Therefore the single screw undergoes a changing load perpendicular to the long axis of the screw, and loosening at the implant-to-bone surface is seen. This results in ‘unstable obstruction’. Loosening of the implant is produced without the load exceeding the strength at the interface. Figure 4c – Diagram of the mechanics of Figure 4b. For a given initial condition the plate pulls and pushes the screw head backwards and forwards. The non-locked screw tilts around an axis within the remote cortex. There is marked resorption where the screw moves while the screw thread near the axis of rotation does not show bone resorption. Similarly, the amount of callus formed corresponds to the amount of displacement or instability.

After conservative treatment the effect of resorption of the fracture surface of the bone is beneficial.⁶¹ The same holds true for flexible fixation. For a given amount of instability resorption increases the distance between the moving surfaces and thus reduces the deformation or ‘strain’ of repair tissues.⁶² The basic working hypothesis of the ‘strain’ theory⁶³ is that a tissue cannot be produced under strain conditions which exceed the elongation at rupture⁶⁴ of the given tissue element, such as a cell. A minimal amount of strain seems to be a precondition of mechanical induction of callus because stably fixed large defects do not show formation of bone.

Compression applied to the fracture produces preloaded continuous contact and thus minimises interfragmentary strain. This enables the osteones to cross the fracture at the compressed surfaces (Fig. 2). The osteones may also cross small gaps which are stabilised by neighbouring compressed areas of contact. However, minimal strain does not induce the formation of callus. After flexible fixation with displacement the net effect of the resorption is widening of the gap (Figs 5 and 6) with consequent reduction of strain to tolerable limits. The mechanism of instability induced bone surface resorption at the fracture gap with the shape of a ‘teardrop’⁶⁵⁻⁶⁷ and allowed tissue differentiation by reducing strain for a given amount of instability. Direct comparison of the finite-element predictions in gaps with the histological findings in osteotomies in sheep showed that the spatial distribution of bone resorption at the ends of the fragments of the fracture directly corresponded to the locations of elevated tissue strain.⁶⁸ The same mechanism of resorption at the interface between implant and bone has a deleterious effect since it increases instability (Figs 3b and

4) which, under poor biological conditions, may not be overcome and will result in delayed or nonunion.

Advantages of instability after biological internal fixation. According to the strain theory an elastic flexible fixation is compatible with the indirect type of healing provided that very small unstable gaps of high strain are avoided. This condition is usually met with indirect reduction by either a bridge plate or a locked nail. The second condition for uneventful healing is that a minimum of biological interference has taken place with avoidance of surgical cleaning of the site of the fracture in an attempt to allow overly precise reduction. In spite of the lack of so-called medial support¹ loosening and/or fatigue of the implant is avoided by the production of early bridging with callus which gives efficient biological medial reinforcement. Two aspects are worth mentioning. According to the strain theory the amount of mobility allowed depends less on the displacement of the fragments alone than on the relation of the width of the fracture gap (L) and displacement (δL); $\epsilon = \delta L/L$. The amount of strain (ϵ i.e. relative deformation) optimally ranges between the minimum required for the induction of callus and the maximum which allows bony bridging. Based on the observations of Hente et al,⁶² the effects of strain may be summarised as follows. The absence of dynamic relative deformation results in lack of mechanical induction of callus formation. Very small amounts of strain induce callus formation. Strain values up to 2% are tolerated by lamellar bone tissue, up to 10% are tolerated by the three-dimensional configuration of woven bone (Fig. 6) and between 10% and 30% induction of resorption prevails.

The situation with strain is different in simple (Figs 7 and 8b) and multiple fractures (Fig. 8d). Biological internal fixa-

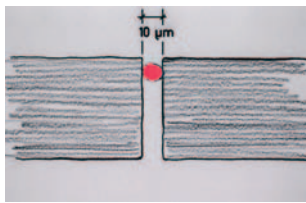


Fig. 5a

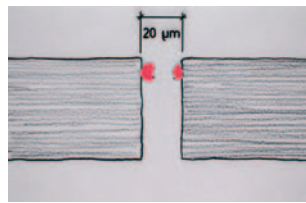


Fig. 5b

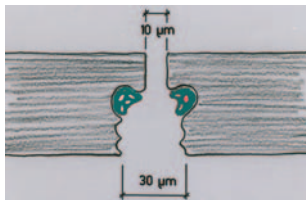


Fig. 5c

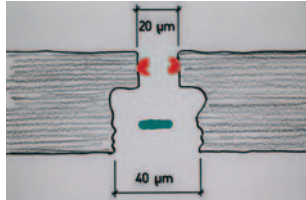


Fig. 5d

Figure 5. The relation of gap width and instability producing strain. Even very slight instability produces high strain conditions within a small fracture gap. Gap and displacement are not visible to the naked eye. Therefore this critical condition largely goes undetected. Figure 5a – A gap of 10 µm with a cell inside. Figure 5b – When the fracture fragments move 10 µm apart, the cell size is increased by 100%, which is the critical value for rupture of the cell.⁵⁰ Figure 5c – It is assumed that part of the fracture gap is widened through osteoclastic resorption to a gap size of 30 µm. Figure 5d – The same displacement as in Figure 5b (+10 µm) causes an increase of the widened part of the gap in Figure 5c, from 30 to 40 µm, that is an increase of only 30%. The cell in the widened gap undergoes tolerable deformation while that in the small gap ruptures or cannot be formed at a strain of 100%. Figure 5e – Photomicrograph of a small gap in the osteotomy of the tibia of a sheep fixed by a splinting plate. The small displacement but high strain prevents tissue differentiation within the small gap while repair tissue fills the widened gap ('teardrop').

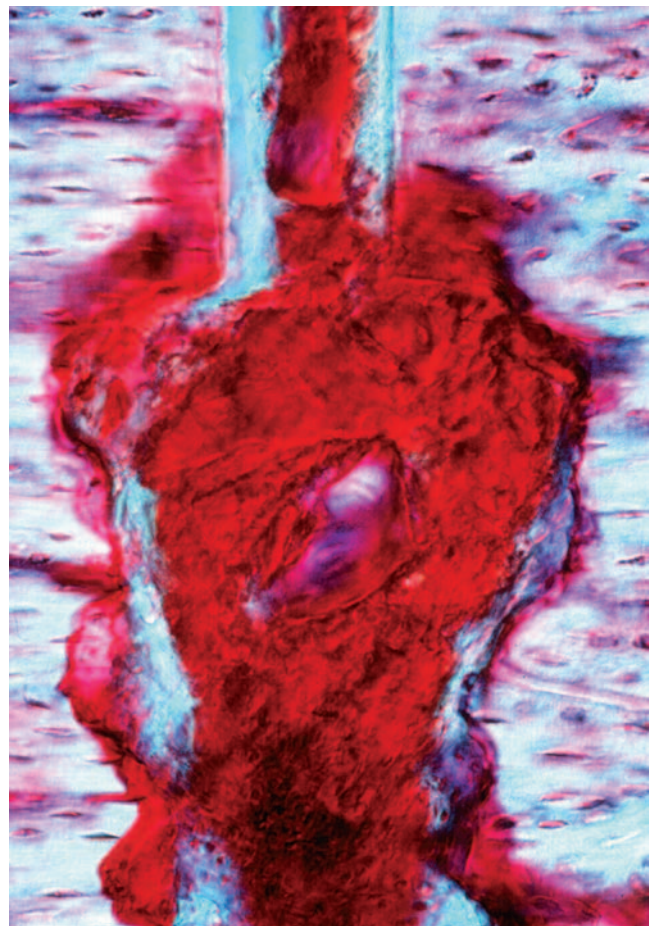


Fig. 5e

tion was first used for multifragmental fractures. Its value in simple fractures seemed questionable. Multiple fracture lines share the displacement and are thus more tolerant to instability while simple fractures must bear the full shift. In order to maintain the strain below critical values it is advisable to undertake a reduction which leaves a larger gap width in simple fractures (Fig. 7).

Means of achieving elastic flexible fixation. This can only be achieved without interfragmentary compression. A splint is a more or less rigid body which reduces but does not eliminate displacement under load. Attempts have been made to use more flexible implant materials such as fibre-reinforced plastics⁶⁹ or carbon plates.⁷⁰ The effect of the dimensions of the implant on its structural bending stiffness is much greater than are changes in Young's modulus. It is more practical to achieve flexibility by reducing the dimension of a metal implant. A combination of a more deformable metal, such as titanium, and slightly reduced size is usually used.

Instability and the risk of corrosion with non-locked screws in flexible fixation. When conventional techniques of internal fixation without compression are used increased fretting may be expected between the elements of a combined

system of screws and plates or nails. It is advisable to use a highly corrosion-resistant material such as titanium.⁷¹⁻⁷⁴ With this material fretting corrosion can be avoided, but fretting abrasion (galling) may occur. In the presence of minimal corrosion the release of soluble metal into the tissue is negligible but particles are released which may form a visible deposit and these may be transported through the lymphatic system.

The major disadvantage of titanium is its limited ductility (plastic deformation before failure), which may be a problem for the expert who limits the amount of tightening of the screw by feeling the 'giving-way' of steel before failure.⁷⁵ This can be solved using the principle of locked screws because there is a sharp increase in torque when the screw locks (Fig. 9). In a tightly monitored prospective clinical study⁷⁶ in which more than 2000 PC-Fix cp titanium screws were applied by 108 different surgeons, no breakage of screws was observed. Elastic flexible fixation may avoid force transmission at the implant-to-bone interface by friction.⁷⁷ It is best achieved by solid anchorage of the screws (locked, threaded bolts) within the plate-like connecting bar of the internal fixator.⁵⁴ To avoid push out and jamming of the locked screws at removal, the initial Morse cone has been

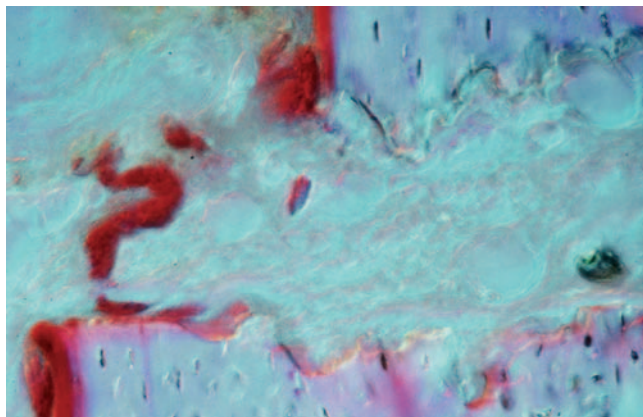


Fig. 6a



Fig. 6b

Photomicrographs of strain-induced resorption of osteotomy surfaces and of initial bridging. The histology is taken from the edge of a transverse osteotomy of a sheep tibia which has been plated. There is small dynamic displacement between the surfaces of the fracture. Because the gap is also small the strain between the opposed surfaces is high. Figure 6a – The opposed surfaces show bone resorption which results in widening of the gap with consequently reduced strain. Between the upper fragment and the outer part of the lower fragment surface (red circle) the distance is larger and the strain smaller. This part of the lower surface does not show resorption of the bone surface. This observation supports the view that such resorption, induced by high strain, reduces the strain to tolerable amounts. When the strain is lower a precursor of callus is seen which will eventually bridge the gap in the shape of a three-dimensional spring. A folded connection (like woven bone) acts as a spring; its elements are less deformed than they would be with a direct connection. Woven bone thus provides improved tolerance to instability. Figure 6b – A modified diagram shows those parts of the initial callus bridge which are missing in this specific histological saw cut. Note the resorption as explained above.

replaced by a conical threaded screw connections (Fig. 9c). A recent development allows achievement of conventional compression as well as locked internal fixation (locked compression plate, LCP)⁷⁸(Fig. 9d). Although the combination of the two (possibly incompatible) modes of application within the same fragment should be the exception.

Potential benefits of avoiding bone necrosis

Avoiding additional damage to the blood supply to bone is an important element in the philosophy of biological internal fixation. The different aspects of bone necrosis will therefore be discussed.

Aetiology of early temporary porosity. The following findings concern exclusively the early temporary porosity seen within cortical bone near implants usually between two and five months after operation. The validity of Wolff's Law⁷⁹ and its more recent interpretation⁸⁰ concerning long-term bone loss under long-lasting conditions of unloading are not challenged.

Earlier it was generally accepted that unloading or stress shielding of the bone caused the bone loss seen near implants which could be expected according to Wolff's Law. This bone loss, a stress protection phenomenon, was first observed in cortical bone deep to plates,^{69,81-86} and also around nails.⁸⁷ The shape of the area of porosity did not correlate with a possible pattern of unloading and the aetiology of the early temporary porosity was therefore reconsidered. A good correlation of the porosity with the width of contact of the implant and hence damage to the blood supply was observed.⁸⁸ Plates with reduced contact with

bone were evaluated and it was seen that there was less porosity with a reduced contact surface.⁸⁹ Furthermore, the use of softer plastic plates which produced less unloading but had a tighter contact between implant and bone did not reduce but enhanced the amount of early temporary porosity.⁹⁰ These observations demonstrate a close link between necrosis and internal remodelling which results in the early temporary porosity seen in the area of necrotic bone.

The different effects of internal remodelling. It was therefore concluded that such early temporary loss of bone was induced by necrotic bone⁸⁸ which stimulates the internal remodelling of the Haversian system within the adjacent living bone. Haversian remodelling progresses through a stage of opening tunnels by the cutter heads (Figs 10b, t1 to t2) resulting in porosity (Fig. 10b, t2). The tunnels are then filled with newly-formed bone (Fig. 10b, t2 to t3) with disappearance of porosity as the area is replaced by living bone.

The observation that bone porosity disappears while the implant is still in place (Fig. 11b) raises further questions about the explanation of temporary porosity on the basis of Wolff's Law.⁷⁹

Formation of sequestra by intensified remodelling. Early temporary porosity by itself may be viewed as a phenomenon without practical consequences because the area undergoing temporary weakening is mechanically protected by the implant. However, when any irritation of the bone such as an infection occurs, the intensified remodelling may lead to the formation of bands of confluent pores. This may result in loss of continuity and the development of sequestra beneath the plate or around a nail,^{91,92} which in turn may facilitate and maintain infection (Fig. 12).^{90,93}



Fig. 7a

Fig. 7b

Fig. 7c

Fig. 7d

Radiographs showing biological internal fixation of a simple spiral fracture of the tibia. Simple fractures can be successfully treated by biological internal fixation. The advantage of this procedure is the formation of early callus bridging and the avoidance of necrosis. An external fixator had been used to reduce the trauma related to an otherwise open reduction. Figure 7a – Radiographs showing an open gap with the internal fixator (LCP) used with locked screws. The splint barely touches the bone because the locked screws allow reliable maintenance of the initial distance between the internal fixator and bone. Figure 7b – Radiograph six weeks after operation showing good progress of union. Figure 7c – Radiograph at six months showing a dense and homogeneous bone structure. Figure 7d – Radiographs at one year.

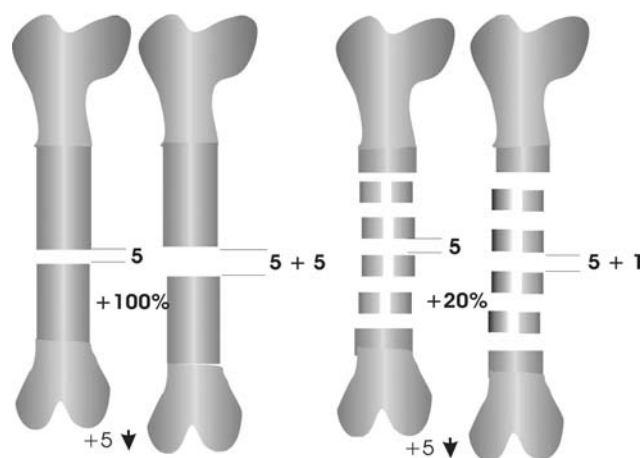


Fig. 8a

Fig. 8b

Fig. 8c

Fig. 8d

The tolerance of instability of simple *versus* multiple fracture lines. The overall displacement is the same (5 mm) between Figures 8a and 8b and Figures 8c and 8d. Figure 8a – Model of a simple transverse fracture with an initial gap width of 5 mm. Figure 8b – The full amount of the displacement of the end fragments (5 mm) is active within the fracture gap, the width of which increases from 5 to 10 mm. This is equal to 100% strain which is the limit of tolerated strain for granulation tissue. Figure 8c – Instead of a simple fracture a multifragmentary fracture with five fracture gaps is seen. Figure 8d – Five gaps share the overall displacement. Thus, each displaces from 5 to 6 mm. The resulting strain is only 20%.

This explains why multifragmentary fractures are more tolerant to instability than simple fracture lines. However, when choosing an initial somewhat larger gap width the strain can be kept low in simple gaps as well. Figure 7 demonstrates that simple fractures can be efficiently treated with elastic flexible fixation.

The practical clinical value of this understanding is that reducing the implant contact provides a feasible alternative, while stress shielding may conflict with stabilisation.

Plate-to-bone contact also depends on the relation of the radius of the curvature of the bone surface to the radius of the undersurface of the plate (Figs 13a to 13c). Figure 13d shows the different areas of contact in relation to the shape of the undercuts of conventional DCP,⁹⁴ LC-DCP⁹⁵ (limited contact DCP), and PC-Fix⁵⁴ (point contact fixator) plates, with the latter showing isolated small points of contact. *Implant contact and bone necrosis.* Once it was accepted that early temporary porosity and implant contact were related, means were sought to reduce the contact of the plate (LC-DCP, see Fig. 13d). Plates depend on friction to transmit forces and a minimum amount of load-carrying surface is required to withstand the contact pressure which produces

friction. The minimum degree of friction produced is about 0.4 times the contact pressure.⁷⁷

Bone necrosis which occurs as a consequence of trauma is not under the control of the surgeon, but it is useful to understand its consequences. Iatrogenic bone necrosis is the result of the surgical approach to bone, direct manipulation to enable reduction and the preparation for application of the implant by reaming of the endosteum and stripping or squeezing of the periosteum. Less obvious, but more important, is the effect of the contact of the implant with the periosteal and endosteal surfaces which carry the blood supply to bone. An implant which contacts bone along an extended surface, such as a plate fitting the radius with a smooth undersurface, inhibits blood reaching or leaving the bone. The observation that the width of the area of so-called stress protection porosity is closely linked to the width of the con-

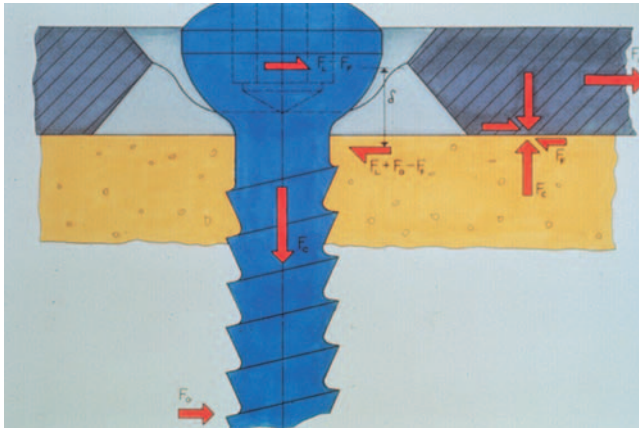


Fig. 9a

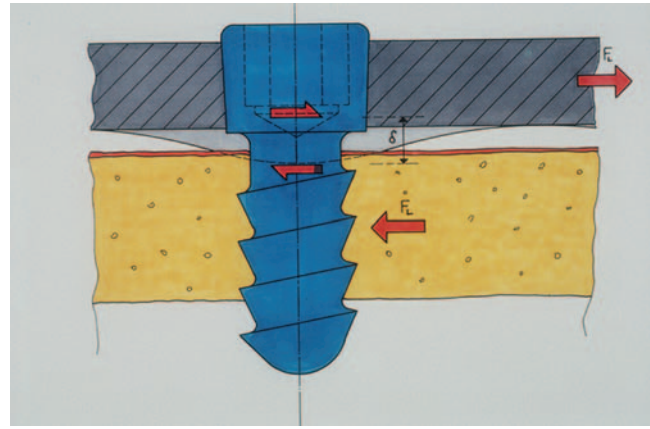


Fig. 9b

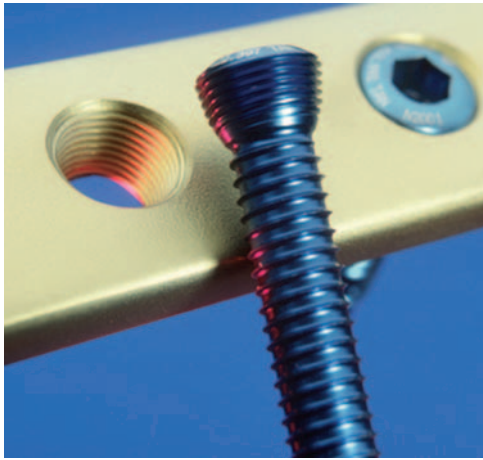


Fig. 9c

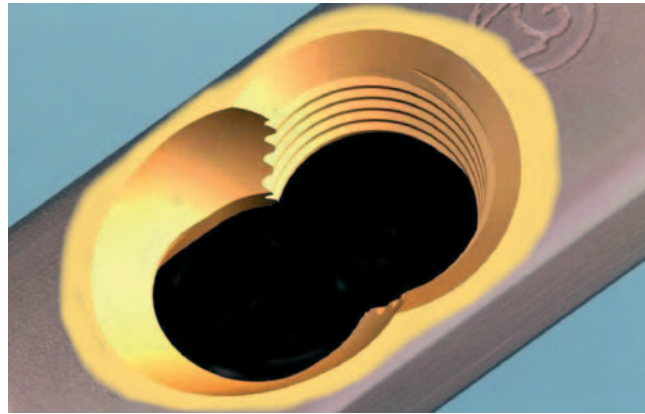


Fig. 9d

Principle of the locked screw. Figure 9a – Longitudinal section through the screw hole of a conventional plate screw. The inclination of the screw is not locked. The screw is tightened with an axial traction of 2000 to 3000 N, the plate is compressed onto the bone producing friction which resists a tangential load of ~1000 N (friction = $0.4 \times 2500 = 1000$ N). Figure 9b – Longitudinal section through the locked screw of an internal fixator. Because of the steep conical surfaces ('Morse cone') the screw locks upon application of minimal torque. Therefore absence of compression between the plate and bone allows either point contact or no contact thus enabling reduction of the contact damage to the blood supply. This type of force transmission does not depend on axial preloading of the screw. Figure 9c – If screws placed as in Figure 9b are tightened with a large torque removal may be difficult. The conically threaded undersurface of the locked screw is more tolerant to excessive torque providing locking and enabling reliable removal of the screw (LISS). Figure 9d – LCP: combination hole allowing either fixation like a conventional DCP or application of locked screws as in the PC-Fix.

tact of the implant supports this view. Observations concerning the area of contact⁹⁶ are explained by the cross-sectional fit between the undersurface of the plate and the contact surface of the bone (Figs 13a to 13c). It should be noted that a trapezoidal cross-section of the plate helps to reduce the area of contact, and also to avoid bone growing over the top of the implant. The inclined lateral surfaces make removal of the implant easier.⁹⁷ Today's plates, such as the LC-DCP,⁹⁵ use a trapezoidal cross-section which varies along the length of the plate.

Under otherwise undisturbed conditions, the mechanical effect of early temporary porosity is minimised since the implant protects the bone. This protection ceases with the removal of the implant. When the situation is compounded

by infection or other causes of irritation, the increased intensity of remodelling may weaken the bone and confluence of the pores may result in the formation of a sequestrum (Fig 12a). This explains why in an infected plate osteosynthesis the sequestrum is prismatic in shape (Fig. 12b) while with an infected nail it appears cylindrical⁹² (Fig. 12c), as does that around a Schanz screw. If early temporary bone porosity is induced by necrosis rather than by unloading, it is possible to reduce the incidence of sequestration by keeping the bone alive by indirect reduction¹² and minimal contact with the implant.⁸⁹

Local resistance to infection and necrosis. Earlier research has shown the effect of stability on susceptibility to infection.⁹⁸ Irrespective of the amount of compression exerted,

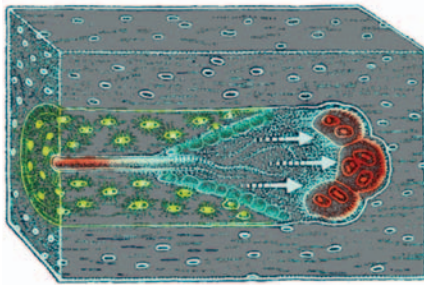


Fig. 10a

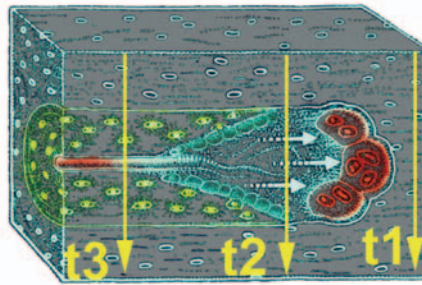


Fig. 10b

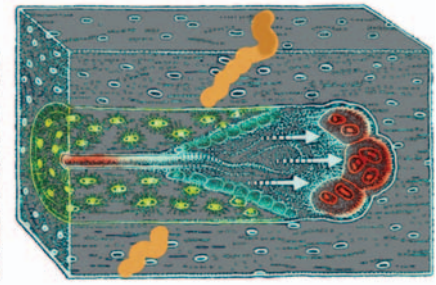


Fig. 10c

The different effects of internal (Haversian) remodelling of cortical bone. The effect of the osteones results from cutter heads tunnelling the bone in front of the osteone and a cone of osteoblasts producing living bone, with osteocytes connected among themselves and to the blood supply. Figure 10a – Diagram showing that the net effect of internal remodelling consists of replacement of dead bone with living bone.⁵⁰ Figure 10b – Diagram showing the different stages of internal remodelling of necrotic bone:

- | | | |
|--|----------|-------------|
| (t1) in front of the cutter heads the bone is dead | → dense | dead bone |
| (t2) during remodelling, behind the cutter heads | → porous | dead bone |
| (t3) after remodelling, behind the osteoblast cone | → dense | living bone |

Internal remodelling goes through a phase of early temporary porosity which is not induced by stress shielding. Adaptation of the cortical bone structure to loading¹⁴ (stress protection) takes years. Figure 10c – Diagram showing remodelling of broken bone. Internal remodelling unites fragments of broken bone. This remodelling, induced by necrosis, restores the continuity of bone within the cross-sectional area of the osteone.

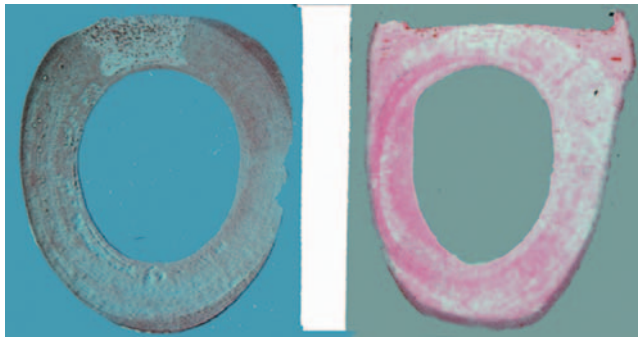


Fig. 11a

Fig. 11b

The aetiology of early temporary porosity. Deep to the plate an area of early temporary porosity develops. Within a few months it fills up with newly-formed bone. The width of the area of porosity corresponds to the width of the contact with the implant. The depth varies. For better visualisation the area of porosity is represented digitally enhanced. Figure 11a – An area of early temporary porosity in a plated sheep tibia at three months. The width of the area of remodelling corresponds to the width of the plate contact. Porosity starts at the borderline of living to necrotic bone and progresses centrifugally towards the implant. An area of refilling of the osteones follows the area of porosity. Figure 11b – A similar cross-section as in Figure 11a but after one year. The area of earlier porosity is refilled by living bone. Along the plate, ramps of callus are seen.

implant contact results in soft-tissue necrosis. Hence the aim must be to reduce the area of necrosis in and near the area of contact which impedes the resistance to infection. Infection may spread along an extended contiguous area of necrosis. Here the aim must be to reduce the area of necrosis and to isolate areas of contact from each other. It may well be that the foreign-body effect which was thought to diminish resistance to infection is due less to the foreign material but more to tissue necrosis and the dead space produced. The use of the principle of internal fixation is to minimise and isolate contact of the implant, and its surface must allow adherence of the soft tissue to avoid a fluid-filled dead space

as is often seen around conventional electropolished steel implants.⁹⁹

The local resistance to infection was studied experimentally using human pathogenic *Staphylococcus aureus* in simulated internal fixation in the tibia of the rabbit. The study observed the effect of the design, material and application of the implant in relation to the number of colony-forming units required to produce an infection. The effect of dead space (slotted *versus* solid nail),¹⁰⁰ the implant material (steel *versus* titanium *versus* degradable polymer),¹⁰¹⁻¹⁰³ the application (reaming, approach)^{104,105} and the design of the implant to minimise contact with bone were evaluated. The overall difference in infection between the groups which utilised a DCP in steel with surface contact and that which had a PC-Fix in titanium with point contact was a ratio of 1:450. When PLA (biodegradable implant material made of polylactide) was used the ratio rose to about 1:7000¹⁰⁶ (Fig. 14).

Refracture and necrosis-induced remodelling. Necrosis immediately deep to an implant may also be related to refracture (Fig. 15). While necrotic bone is roughly similar in strength to living bone the situation at the site of the fracture is different. Necrosis of the bone immediately deep to an implant will result in local delay or impairment of internal remodelling. The resulting lack of healing may facilitate refracture. This effect is the more harmful when, according to conventional technology, the plates are placed at the side of the bone where functional loading produces tension. After removal of the plate, the local lack of healing can act as a stress riser¹⁰⁷ and result in a refracture caused by traction induced through bending.

Consequences for surgical technology

The main requirements are avoidance of surgical exposure, the minimisation of damage due to instrumentation and the

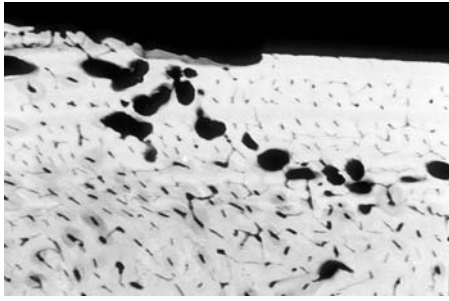


Fig. 12a



Fig. 12b

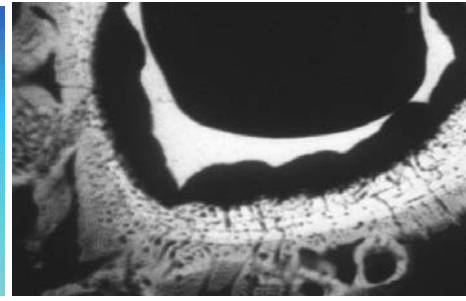


Fig. 12c

Adverse effect of intensified early temporary porosity. As in the preceding figure the porosity is located between the living and the necrotic bone in close contact with the living bone. The figures show the relation between intensified porosity and possible production of a sequestrum. Figure 12a – Deep to the plate an area of intensified porosity has developed. When the pores merge a sequestrum is produced. Figure 12b – The relationship between the cross-section and the 3-D aspect. A prismatic sequestrum has separated. Figure 12c – Intensified remodelling and formation of a ring sequestrum around the nail.

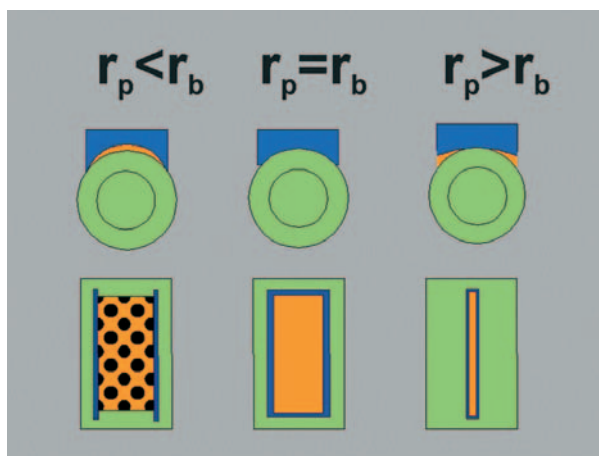


Fig. 13a

Fig. 13b

Fig. 13c

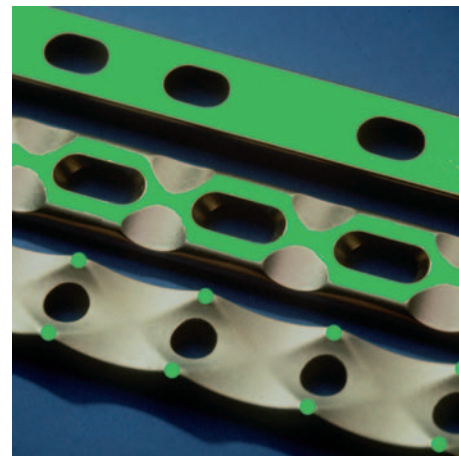


Fig. 13d

Reduction of contact of a plate to bone can be achieved by appropriate selection of the radius of the plate undersurface or by undercutting the latter. The minimum mechanically tolerated contact area depends on the compression exerted by the screws in relation to the compressive strength of the bone. In addition to the fit on the cross-sectional plane that along the long axis of the bone must be taken into account. Figure 13a – When the radius of the plate undersurface is smaller, a two-line contact results corresponding to the width of the plate. Such contact impairs the blood supply similarly but somewhat less than is shown in Figure 13b. Figure 13b – Identical radius of curvature results in maximal surface of snug contact. Figure 13c – When the radius of the bone surface is smaller than that of the plate, the width of the contact decreases and may be a line only. The disadvantage of such a solution is that it provides adequate resistance to torque only with locked bicortical screws. Figure 13d – The area of possible maximal contact depends on the shape of the undercuts (above, conventional DCP; middle, LC-DCP; below, PC-Fix).

reduction or elimination of implant contact. There have been several developments with low contact plates and fixators.^{21,108,109} In external fixators the transcortical pathway for infection offsets the positive effect of minimising implant contact to bone and flexible fixation.

Methods of improving implant anchorage in porotic bone. Porotic bone presents an increasingly important problem in the surgical treatment of fractures. Several methods may be used to improve anchorage of the implant.

1) While in conventional plate fixation axial traction per plate screw of 2000 to 3000 N is usually applied,¹¹⁰ part of this force, otherwise used to maintain compression of the interface preload, improves the pull-out strength of locked screws.

2) Locked screws which are applied inclined in relation to each other (Fig. 16) improve the anchorage because more bone must be displaced than with parallel or non-locked screws when stripping is to occur. Long locked screws are preferred. Non-locked screws have been used in the inclined position for a different purpose when they were tilted mainly to improve their function as lagged plate screws.¹¹¹

3) The area of contact must be able to bear the pressure load of conventional plate fixation. Point contact cannot be realised when the plate screws cause compression of the undersurface of the interface plate to bone.

4) It is important to consider the effect of the stiffness of the implant (see Fig. 17a) in respect of internal fixation of porotic bone. Fracture healing which tolerates some instability allows the use of flexible implants, which produce less

Table I. Recovery of strength in fractures treated by implants with extensive contact between the undersurface of the plate and bone surface (DCP) compared with implants with minimal point contact (PC-Fix). At 12, 24, 48 and 96 weeks the implants were removed and the bending strength was assessed as a percentage of the opposite intact bone

Weeks	DCP (n = 6)		PC-Fix (n = 6)		p value of difference*
	Mean	SEM	Mean	SEM	
12	45.8	1.7	64.5	2.1	<0.001
24	73.8	4.6	68.5	3.6	NS
48	73.3	0.9	78.7	5.2	NS
96	56.7	2.2	68.8	2.6	0.003

*Student's t-test

pull-out for the same deflection. This effect must be weighed against tolerance to instability. To achieve the same amount of strain, however, the gap width may be adjusted.

5) The thread of the screw may contain a thin or thick core. The pull-out strength of the screw depends mainly on the outer diameter. Therefore, a thicker core, with better purchase against forces acting perpendicular to the long axis of the screw, may offer better anchorage. The thicker core also increases bending stiffness, which is important in long locked screws.

The strength of internal fixation may be enhanced by using long implants (Fig. 17b). For a given amount of bending moment the pull-out force depends inversely on the distance between the fulcrum at the surface of the fracture and the position of the screw.

The internal fixator using the Point Contact principle (PC-Fix). To study the concept of the internal fixator the Point Contact Fixator (PC-Fix) has been developed.^{54,112} It handles like a plate but acts like an internal fixator. The PC-Fix was used to study the concept of the internal fixator with unicortical screws as required by minimally invasive percutaneous osteosynthesis (MIPO). It may also be used with bicortical screws in metaphyseal or porotic bone.

The strength of fracture healing¹¹³ is of interest in determining when the implant can safely be removed. In experimental studies using short oblique fractures in the tibia of the sheep (Fig. 18) an astonishingly early recovery of strength was observed in fracture healing (Tables I and II).¹¹³⁻¹¹⁶

Table II. Regaining strength of healing with conventional DCP versus PC-Fix treatment at 12, 24, 48 and 96 weeks.¹¹⁶ The location of the occurrence of bone breakage while testing is listed. When the healing fracture failed within the original fracture gap, this symbol used is ReFrX. When the original fracture did not fail, but the new fracture occurred outside (mostly through the next screw hole) NoReFrX is used. When the fracture union is stronger than its surrounding bone, including the former screw holes, removal of the implant can usually take place

12 weeks		24 weeks		48 weeks		96 weeks	
DCP	PC-Fix	DCP	PC-Fix	DCP	PC-Fix	DCP	PC-Fix
ReFrX	NoReFrX	NoReFrX	NoReFrX	ReFrX	NoReFrX	NoReFrX	NoReFrX
ReFrX	NoReFrX	ReFrX	NoReFrX	NoReFrX	NoReFrX	NoReFrX	NoReFrX
ReFrX	NoReFrX	NoReFrX	NoReFrX	ReFrX	NoReFrX	NoReFrX	NoReFrX
ReFrX	NoReFrX	ReFrX	NoReFrX	NoReFrX	NoReFrX	NoReFrX	NoReFrX
ReFrX	NoReFrX	NoReFrX	NoReFrX	NoReFrX	NoReFrX	NoReFrX	NoReFrX
ReFrX	NoReFrX	NoReFrX	NoReFrX	NoReFrX	NoReFrX	ReFrX	ReFrX
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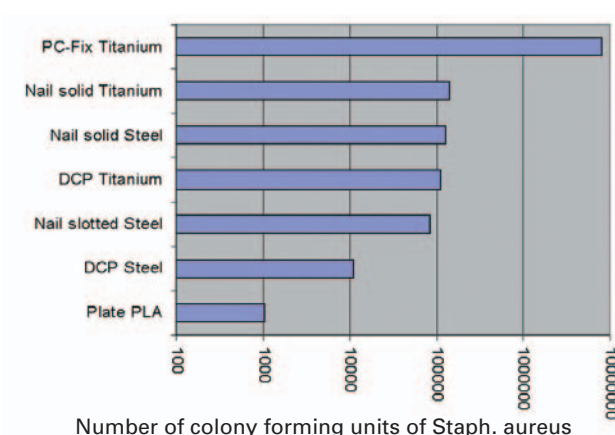


Fig. 14

Local resistance to infection. The number of colony-forming units of *Staphylococcus aureus* which results in an incidence of 50% of infection is plotted on the abscissa. The different designs of implant, materials and application are grouped on the ordinate. Material, design and mode of application play a role.

Testing the PC-Fix principle clinically in animals. When screws are locked they cannot tilt. Since they are under minimal tensile preload they do not need to be anchored in two cortices. The locking within the device replaces the stabilising effect of the second cortex. The minimal traction exerted along the screw-bolt axis means that increased holding force can be expected under certain preconditions. To study the performance of unicortical screws under difficult conditions, clinical testing with the PC-Fix was performed in routine veterinary surgery. In small animals, the internal fixator was removed between 12 and 18 weeks without pull-out or refracture under unrestricted weight-bearing.¹¹⁸ The same principle was also tested in large animals.²⁵

Clinical testing of the PC-Fix principle in man. In a prospective series of 387 consecutive cases with follow-up of 97% the results were encouraging and proved the validity of the concept.⁷⁶ Further clinical studies confirmed the findings.^{117,119} This method of achieving stabilisation of the fracture does not depend on extended or confluent implant contact. It allows handling and results in low rates of infec-

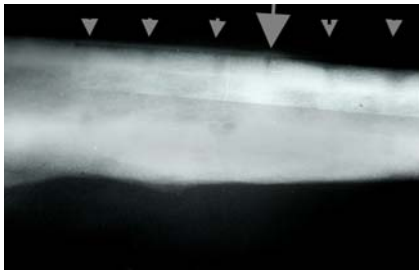


Fig. 15a

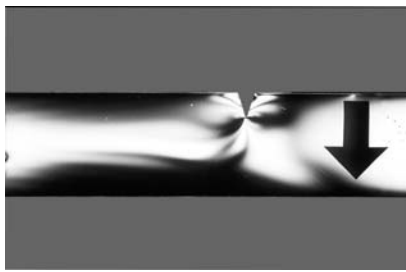


Fig. 15b

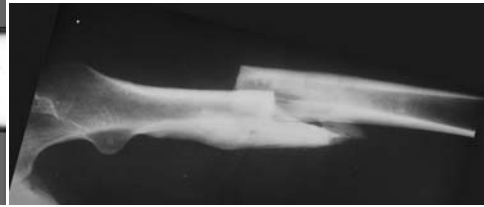


Fig. 15c

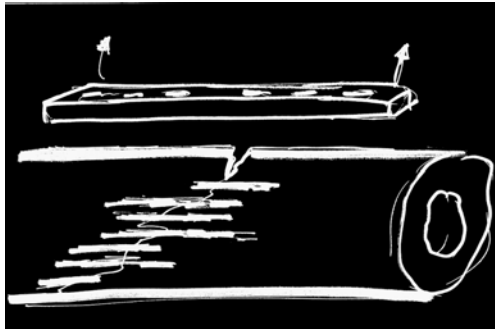


Fig. 15d

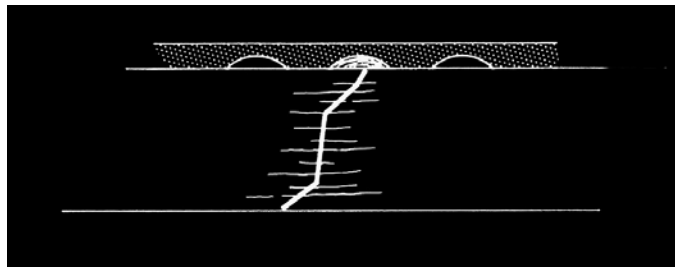


Fig. 15e

Refracture and plate contact. Damaged blood supply near the plate resulting in a refracture after removal of the plate. Figure 15a – Radiograph of a femur after removal of the plate. The large arrow points to a small defect immediately deep to the plate. The small arrows indicate the former position of the screws. Figure 15b – A defect of healing as observed in Figure 15a represents mechanically a 'notch' which is a discontinuity at the surface. The photoelastic technique shows that such a notch acts as a stress riser. Figure 15c – At the area of the stress riser a refracture has occurred. Figure 15d – Diagram of the effect of damaged blood supply due to implant contact. In this case, adjacent to the snugly fitting plate, healing near the surface of the bone is retarded or impeded. A notch acting as a stress riser results because the surface of the bone near the former position of the plate is generally under tension. Figure 15e – Diagram of the potential of implants not in contact with the bone surface. In this case of an undercut plate (LC-DCP) the remodelling as a basis of healing is improved and there is space for a small bridge of callus across the fracture.

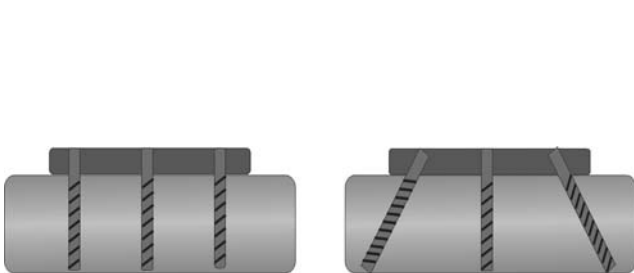


Fig. 16a

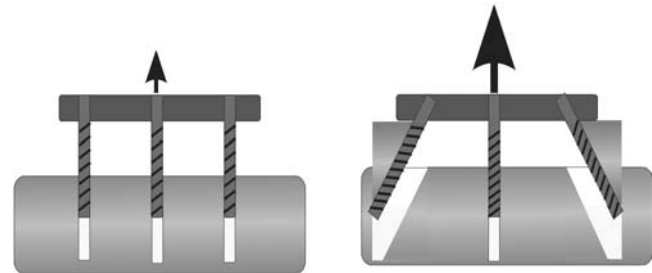


Fig. 16b

Diagrams showing improved anchorage of divergent locked screws. Figure 16a – Before application of the load. Figure 16b – After a pull-out load (e.g. induced by bending) has been applied the width of the area of bone which is involved is larger in the case of locked divergent inclination of the screws.

tion in the rabbit.¹⁰⁵ The device looks like a plate but acts like a fixator without transcutaneous pathways.

Minimally invasive percutaneous osteosynthesis (MIPO).¹²⁰ The internal fixator has similarities to the external fixator in that it touches the bone surface with a minimal area of contact and causes little additional damage to the blood supply. It does not require shaping to fit the bone surface. The connecting rod may stand off without bone contact thus compensating for differences in shape between the

bone and the body of the fixator. This in turn allows the use of an aiming handle which maintains congruency with the implant. It is therefore possible to insert the internal fixator through a small incision remote to the site of the fracture with blind application of the self-drilling screws.

The latter are easy to apply but do not allow determination of the required length of the screw. Therefore, such implants are used preferably as unicortical screws where the tip protrudes into the medullary cavity (Fig. 18b). The use

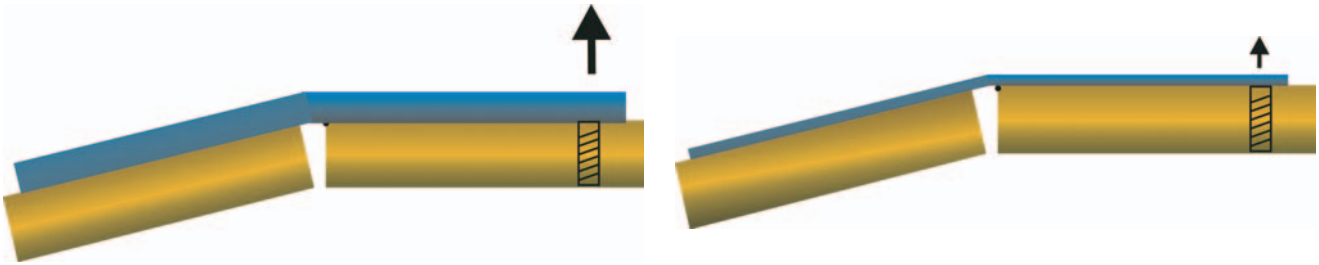


Fig. 17a

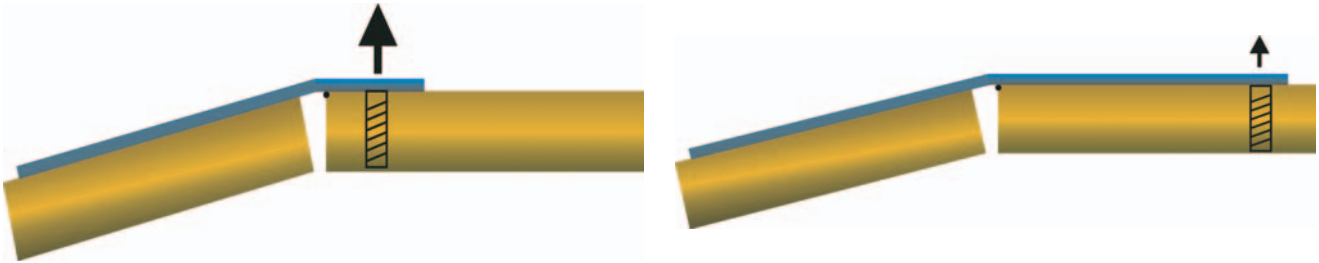


Fig. 17b

Flexibility and length of plate and pull-out load. Figure 17a – For the same amount of deflection a more flexible (thinner) plate produces less pull-out force. Figure 17b – For a given amount of bending moment a long plate produces markedly less pull-out force.

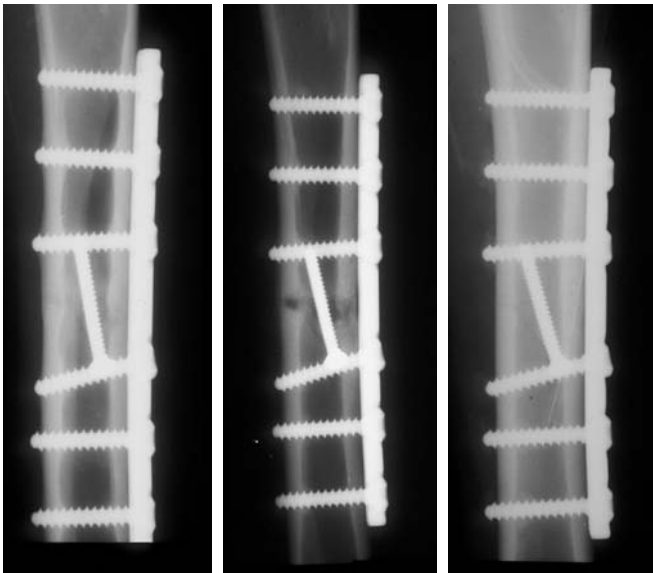


Fig. 18a

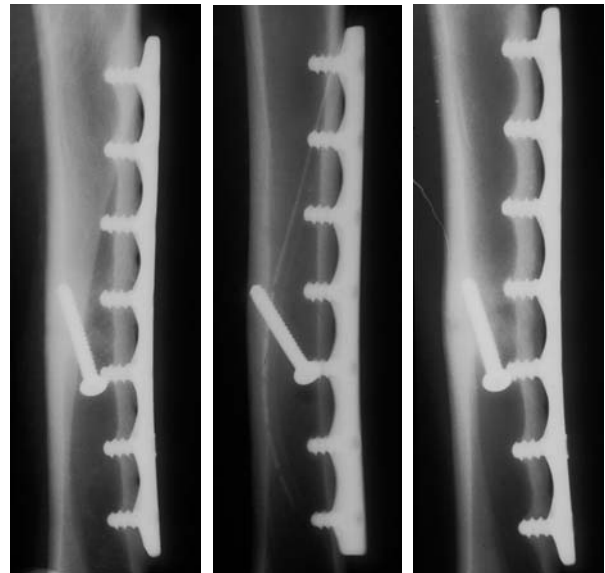


Fig. 18b

Internal fixator and bone healing. A short oblique fracture treated by either a) a conventional DCP or b) with PC-Fix using a free lag screw. The unicortical screw proved to offer enough strength of anchorage. The bone immediately deep to this implant is dense and homogeneous.

of MIPO avoids a large surgical approach and allows the treatment of fractures with contused skin in which the remote skin incision should be an advantage. The experimental data need further clarification.¹²¹

Since no tension-band principle is applied as in compression plating, the internal fixator may be placed on any bone surface which can be conveniently approached. MIPO requires

skill to place the fixator exactly along the bone surface and it is advisable to practice the open MIPO technique first.

Unicortical screws should be employed preferably because the self-drilling screw can be used with a very simple technique of insertion of the screw with the drill.

Internal fixators in current clinical use. The PC-Fix served to prove the concept of internal fixation with locked unicor-

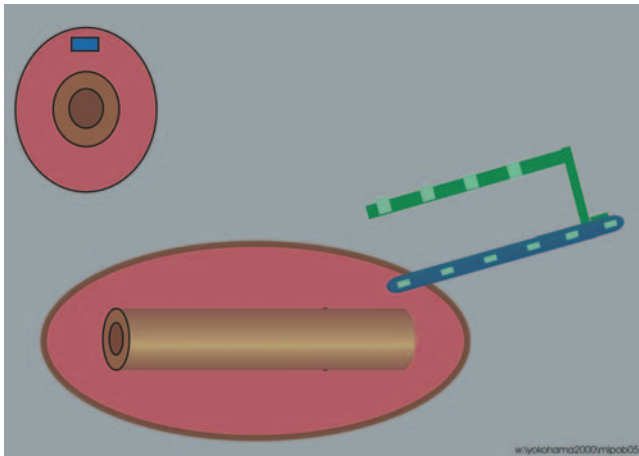


Fig. 19a

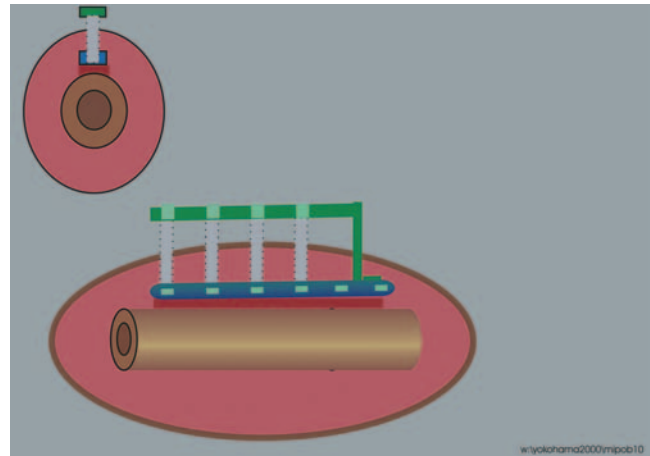


Fig. 19b

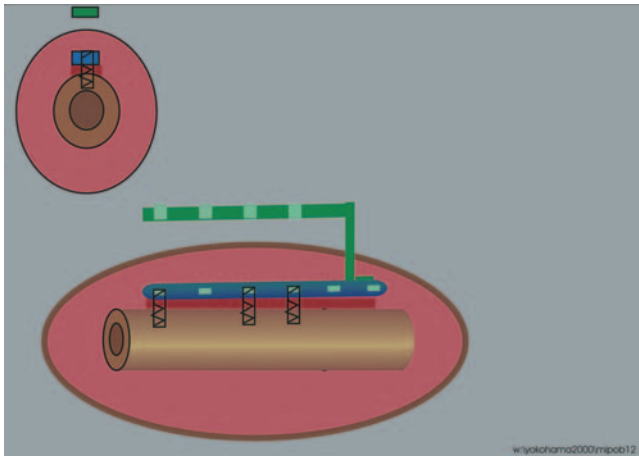


Fig. 19c

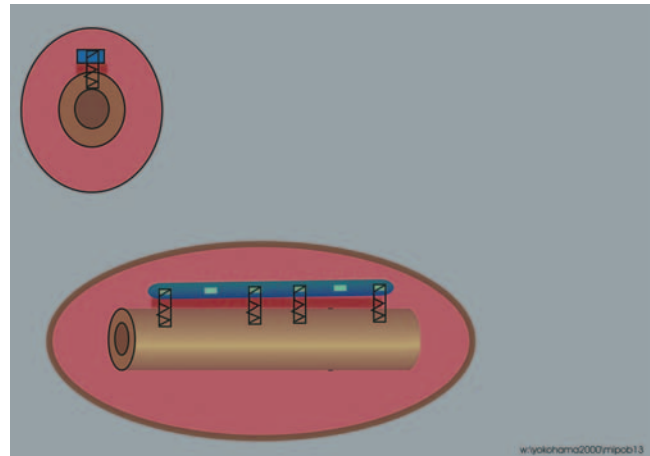


Fig. 19d

Minimally invasive percutaneous osteosynthesis (MIPO). Figure 19a – The insertion of the internal fixator using an aiming handle. Figure 19b – The aiming handle allows the surgeon to find the screw holes within the internal fixator reliably, provided that the body of the internal fixator has not been shaped during surgery. Such shaping is not required as with the external fixator. In both techniques the difference between the shape of the bone and (preshaped) fixator is taken up by differences in the free length of the threaded bolts (screw). Figure 19c – The blind application of self-drilling self-tapping unicortical screws. Figure 19d – Final stage after removal of the aiming handle.

tical screws¹²² and further developments are now in clinical use. The LISS^{123,124} is an internal fixator taking advantage of locked full-length metaphyseal screws, and a combined plate allowing for compression fixation and/or locked internal fixation, the LCP, is in clinical use⁷⁸ (Figs 7 and 9d).

The technique of internal fixation as realised with the PC-Fix,⁵⁴ has been shown to reduce the incidence of infection and to facilitate early solid union in animal studies.^{105,116} The advantages of biological internal fixation are the simplicity of handling, prompt contribution of the bone to healing and resistance to infection and, possibly, to refracture. However, it is difficult to judge if bone is viable and it is important to know this because a combination of necrosis and instability of fixation may cause problems. Internal fixators are at a disadvantage since secondary correction is not possible.

The technology of MIPO would be more attractive if the reduction of the bone fragments which carry joint cartilage could be achieved by computer-aided surgery¹²⁵ or simple mechanical means, since treatment of a fracture would then become simpler, faster and safer.

Differential indications for conventional *versus* biological internal fixation

When the blood supply to the fracture is severely damaged and the bone is necrotic recovery takes many months. Conventional compression fixation then allows for long-term protected internal remodelling. When the blood supply is good, or can be restored within bridges between the soft tissues and bone, biological internal fixation is considered to be the method of choice. The two

methods of stabilisation cannot be equally applicable to the same fracture site. Based on the strain theory, either absolute stability or elastic flexible fixation of a gap is required. Thus, combinations of different techniques must be carefully considered.

Further development

In order to gain full advantage of biological internal fixation and of MIPO, simple methods need to be devised to allow reduction of the main end-fragments. As in locked nailing, biological internal fixation requires only reduction of the main fragments which carry the articular surfaces. A method of reduction and temporary maintenance of the main fragments in a proper three-dimensional position of bending, torsion and length is necessary. Fixation using MIPO with unicortical self-drilling locked screws would then be simple. These aims should be obtainable using computer-aided technology or simple mechanical means.

A method of determining the viability of the bone before or at least during surgery would help to select the method of stabilisation and to improve the prognosis. The exact boundary conditions of strain in respect of amplitude and timing need further study.

With the increasing incidence of osteoporosis the stabilisation of fractures in patients with soft bones is a priority and fracture implants which allow monitoring of the load in vivo will help.^{126,127} The boundary conditions of flexible fixation, namely the limits of strain under clinical conditions, must be further analysed.

An implant material must fulfil many requirements to offer optimal performance. Earlier designs needed to offer high strength and good plastic deformation to allow shaping. These were, to a large degree, mutually exclusive requirements. The technology of internal fixation does not require shaping of an implant body to conform to the surface configuration of the bone during surgery. Therefore, a new field of possible materials including high-strength metals with low ductility or fibre-reinforced mainly non-degradable plastics may now be considered.¹²⁸⁻¹³⁰

Fillers or bone substitutes may provide strength but their absorption and influence on the blood supply need clarification. Open porous, expandable resorbable foams with elastic properties may prove to be useful.

Bone stimulation is most needed in chronic infected non-union, but should avoid killing the cells which are exhausted. In fresh fractures natural stimulation is readily available. The use of bone stimulation in defects is of practical interest but there is strong competition by the ingeniously simple and reliable method of Ilizarov.⁵²

Summary

The advent of 'biological internal fixation' is an important development in the surgical management of fractures.

Locked nailing has demonstrated that flexible fixation without precise reduction results in reliable healing. While external fixators are mainly used today to provide temporary fixation in fractures after severe injury, the internal fixator offers flexible fixation, maintaining the advantages of the external fixator but allowing long-term treatment. The internal fixator resembles a plate but functions differently. It is based on pure splinting rather than compression. The resulting flexible stabilisation induces the formation of callus. With the use of locked threaded bolts, the application of the internal fixator foregoes the need of adaptation of the shape of the splint to that of the bone during surgery. Thus, it is possible to apply the internal fixator as a minimally invasive percutaneous osteosynthesis (MIPO).

Minimal surgical trauma and flexible fixation allow prompt healing when the blood supply to bone is maintained or can be restored early. The scientific basis of the fixation and function of these new implants has been reviewed. The biomechanical aspects principally address the degree of instability which may be tolerated by fracture healing under different biological conditions. Fractures may heal spontaneously in spite of gross instability while minimal, even non-visible, instability may be deleterious for rigidly fixed small fracture gaps. The theory of strain offers an explanation for the maximum instability which will be tolerated and the minimal degree required for induction of callus formation. The biological aspects of damage to the blood supply, necrosis and temporary porosity explain the importance of avoiding extensive contact of the implant with bone. The phenomenon of bone loss and stress protection has a biological rather than a mechanical explanation. The same mechanism of necrosis-induced internal remodeling may explain the basic process of direct healing.

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