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INTRAMEDULLARY FIXATION TECHNIQUES

Part of "6 - PRINCIPLES OF INTERNAL FIXATION"

History

Stimson (218) refers to techniques of intramedullary fixation in which ivory pegs were jammed in the medullary canal in his 1883 textbook on fracture care. Hey-Groves (101) probably inserted the first metallic intramedullary device in a gunshot fracture of the femur during World War I. Writing in 1921 about ununited fractures, he stated:

> It occurred to me, therefore, to use a long internal peg or strut, such as would render unnecessary any further fixation and would afford absolute rigidity. I have used pegs of various shapes, cylindrical, cross-sectional, and solid rods; and I am inclined to think that the last named are the best, because they give maximum strength and there is an avoidance of hollows and crevices which form dead spaces.

Smith-Petersen (175) applied a percutaneous medullary fixation technique to solve immobility problems associated with hip fractures in older patients. Küntscher (130,131 and 132), impressed by the work of Smith-Petersen, conducted basic and clinical research on medullary surgery and fixation, ushering in the modern

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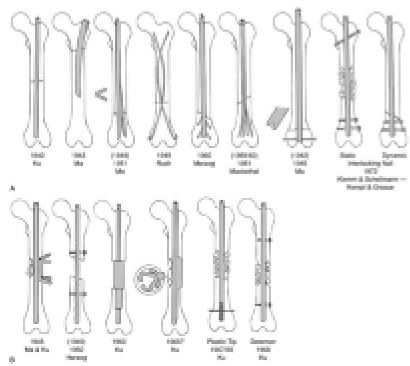
era of medullary fixation. Knowledge of Küntscher's methods spread to Germany's neighbors during World War II as prisoners of war returned home ambulatory after medullary nailing of their femoral fractures (214). Techniques developed for use in the femur were applied to other long bones and several systems of fixation evolved.

Development and Expansion

Since Küntscher's first report in 1939, intramedullary surgery has expanded considerably with respect to new implants, indications, and techniques (13). Maatz et al. (143), pioneers of intramedullary surgery and contemporaries of Küntscher, wrote a fine article on the subject, identifying a wide range of nail designs (Fig. 6-47). The emphasis has been on complete nail systems for reamed and unreamed applications in the femur, tibia, and humerus (Fig. 6-48). Low risk of pseudarthrosis and infection, shortened hospital stay, and rapid return to function have lent further impetus to the extension and application of these techniques to other long bones.

FIGURE 6-47. Medullary fixation has stimulated the



creative instincts of orthopaedists and traumatologists. **A:** Tactics to control rotational instability only. **B:** Tactics to control length or rotational stability by means of blocking. Ku, Küntscher; Ma, Maatz. (Modified from Maatz R, Lentz W, Arens W, et al., eds. *Intramedullary nailing and other intramedullary osteosynthesis*. Philadelphia: WB Saunders, 1986:253–254.)

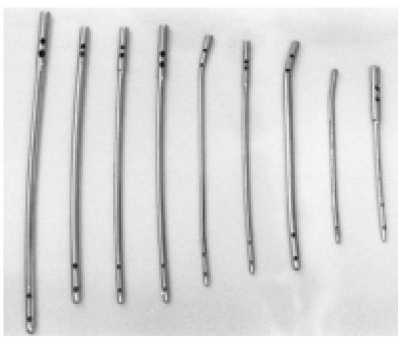


FIGURE 6-48. The Russell Taylor complete nail system. From *left* to *right,* femoral reconstruction nail, femoral nail (after reaming), delta femoral nail (without reaming), delta reconstruction nail (small diameters), delta tibial nail, delta II femoral tibial, humeral, and delta II reconstruction nails. (Courtesy of Smith & Nephew Richards, Inc. Memphis, TN.)

Device Designs

Length, Width, and Curvature

Length, width, and curvature are design features typically matched to the bone in question. Rush rods and Ender pins have small diameters relative to the medullary canal and have uniform curvature but variable lengths. Stability is achieved with stacking of multiple devices within the canal or by using the

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elastic properties of the rod to create a spring-like mechanism (Fig. 6-49) (111,142,158,167,186,197,235). Hackethal developed a method sometimes referred to as bundle nailing, in which as many small rods as possible are placed in the medullary space, eventually creating stability by incremental jamming (68,182).

FIGURE 6-49. Upper **(A)** and lower **(B)** extremity long bone fractures, fixed using Rush techniques and implants. (From Rush LV. *Atlas of Rush pin technics,* vol 243. Meridian, MS: Berivon, 1976.)

More complicated devices have curves to aid insertion, removal, and stability or to duplicate the natural curvature of the bone. Femoral nails are often curved in a sagittal plane to simulate the sagittal bow of the femur. Küntscher (132) appreciated the double sagittal curve of the femur and experimented with a nail having first a posterior curve and then an anterior curve that he referred to as the “dopple nagel” or double nail. Passing a nail with a double curve through a bone having a double curve proved difficult, and the design was abandoned early. Larger and more rigid tibial and humeral nails have been designed that have curvatures to accommodate insertion portals offset from the center of the medullary canal.

Too much curvature built into the nail makes nail removal difficult and hazardous, as was discovered with the Zickel subtrochanteric device. This device has an anteversion, valgus, and anterior curvature to fit the curves of the femur. Once the fracture has healed, the bone remodels to the nail and attempted removal has led to refracture.

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Cross-Sectional Geometry

Intramedullary devices can be solid or hollow; open-sectioned (slotted) or closed-sectioned; and cylindrical, rectangular, diamond, square, cloverleaf, triflanged, or otherwise configured. Solid nails are suitable for placement without canal preparation by reaming but are hard to remove if broken. Hollow nails make insertion over a guide rod possible and are ideal after canal preparation by reaming. The wall thickness of hollow nails may be variable to alter the strength and stiffness of the device. Similarly, a slot may be placed in the device to increase torsional flexibility. Cross-sectional shape influences the mechanical properties of the nail and also has an effect on the return of circulation after nailing. Channels in the nail allow better revascularization than a nail that completely fills the medullary cavity out to the endosteal cortex (191). Flutes and corners on nails have given way to locking with screws for torsional and length stability.

Locking Capacity and Configuration

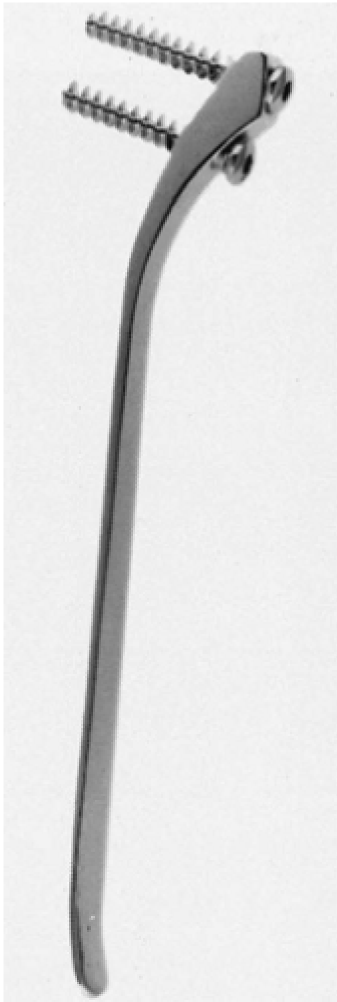
Most contemporary internal fixation devices are designed to allow cross-locking at both ends, so that screws can be placed through the bone and the nail above and below the fracture for additional stabilization. The number, location, and angle of screws may be varied for specific advantages. Distal and proximal location of screws has extended indications of nailing. Oblique screw orientation, common in femoral and humeral nails

(17,58,119,120,124,198), reduces the rotational moment about the screw, eliminating the need for a second screw. Two parallel screws perpendicular to the long axis of the nail are used for distal locking in tibial and femoral nails. The ASIF tibial nail has the possibility of one its three distal screws being oriented 90 degrees to the other two, extending the indications to include more distal fractures (160).

The Huckstep intramedullary compression nail (106,107 and 108) differs from other nails in its quadrilateral cross section and titanium alloy composition. Holes along the length of the nail accept interlocking screws at multiple levels, and oblique proximal holes allow the insertion of lag screws up the femoral neck. Intraoperative radiographs are not routinely required, and because the design is not amenable to closed nailing techniques, open reduction is mandatory.

Proximal locking in the upper femur has been extensively modified to make nailing applicable to proximal femoral fractures from the subtrochanteric region proximally. Initial fixation systems consisted of an implant driven into the head and neck, with a Küntscher nail inserted through the head–neck piece (98,143). Even though the fixation principle has proved to be sound, the “Y” nail of Küntscher had limited popularity, perhaps due to technical difficulties in the insertion technique. Subsequent designs have modified proximal locking, with use of a triflanged nail (Zickel) (51,157) placed thorough the shaft nail, a “U” nail with shaft nail inside (Williams) (157), and one or two screws placed through the nail (Russell-Taylor and others) (23) (Fig. 6-50).

FIGURE 6-50. The Mouradian intramedullary device is used for complex proximal humeral fractures in which the head is unstable relative to the shaft. The lag screws fix the head to the nail while the nail jams in the shaft. Tuberosity fragments are then repaired to the stable head shaft construct with sutures. (Courtesy of Howmedica, Rutherford, NJ.)



Materials

Alloys that are used for nails include 316L steel, 22:13:5 steel, titanium, titanium aluminum vanadium (Ti-6Al-4V), and titanium aluminum niobium (Ti-6Al-7Nb). Except for small-diameter nails used without reaming, differences in material properties are probably less significant in terms of fracture healing than nail diameter and wall thickness or biologic viability of bone at the fracture site. The smaller diameter nails, however, should be made of alloys having superior material properties.

Mechanics of Intramedullary Implants

During the period of fracture healing, internal fixation aids in transmission of forces from one end of the fractured bone to the other, thereby producing stresses in the implant. The mechanical behavior of the implant is determined by both material and geometry. The rigidity or stiffness of a cylindrical structure in bending and torsion is proportional to the fourth power of the radius (i.e., the polar moment of inertia). The farther that material is distributed from the bending or torsional axis, the stiffer the structure becomes. A 1-mm increase in the diameter of an intramedullary nail enhances its stiffness by 30% to 45% (3), and a 25% increase in nail diameter doubles its bending strength (134).

The Küntscher nail had a cloverleaf cross section with a longitudinal slot. Küntscher believed that the open section would allow compression of the nail by the isthmus of the medullary

canal, thereby providing greater rotational control. Placing the slot anteriorly on the tension side of the fracture provides the strongest configuration. When the slot is placed on the compression side, local buckling occurs with high bending loads (4). An open section has little effect on the bending stiffness of a nail but markedly reduces its stiffness in torsion. In a thin-walled cylinder, the addition of a narrow longitudinal slot reduces the torsional moment of inertia to 15% of its initial value (4,135,220).

The working length of a nail is that portion of the nail that spans the fracture site between areas of fixation in the proximal and distal fragments (i.e., the unsupported segment of the nail). This may vary from 1 to 2 mm in a transverse fracture at the isthmus to several centimeters in a comminuted diaphyseal fracture. In a comminuted fracture fixed by a static-locked nail, the working length is the distance between the proximal and distal locking screws. The working length influences nail rigidity in both bending and torsion. In bending, the stiffness is inversely proportional to the square of the working length. A nail having a working length of 0.25 inches is 16 times more rigid in bending than a nail with a working length of 1 inch. In torsion, the stiffness is inversely proportional to the working length, so that doubling the working length halves the torsional rigidity (7). A short working length, therefore, improves nail rigidity both in bending and in torsion.

With torsional loading, a nail both twists and slips within the medullary canal. Slipping allows residual angular displacement after the load is released. Gripping strength is the resistance to slipping at the implant–bone interface and is essential for the transmission of torque between fracture fragments. Cortical reaming to increase the length of cortical contact or by the addition of flutes can increase grip (3). Interlocking nails optimize grip by rigidly affixing the nail to the bone with screws. Kyle and associates (134,135) have quantified gripping strength as the spring-back angle after *in vitro* torsional testing of various nail systems. Spring-back angle is calculated by twisting a bone–nail construct through an arc of 10 degrees and then releasing the deforming force. The final angle measured using the initial starting point as the reference plane subtracted from 10 degrees yields the spring-back angle. The spring-back angle depends on the working length of the implant, the mode of locking (if any), and the torsional rigidity of the implant. Screw fixation proved more effective than fins. As working length is increased, torsional stiffness decreases with a small increase in spring-back angle, creating a spring-like action (135).

Johnson and Tencer et al. (114,223,224 and 225) have conducted extensive *in vitro* biomechanical evaluation of simulated comminuted subtrochanteric and femoral shaft fractures fixed by a variety of intramedullary nails. In three-point bending, fracture models having segmental subtrochanteric defects fixed with interlocking nails were 55% to 70% as stiff as intact femurs, and fractures fixed with Ender nails had less than 25% of the bending stiffness of an intact femur. Femurs with segmental defects of the shaft were significantly less rigid in bending for all intramedullary devices tested.

Models tested in axial loading showed wide variations. Ender nails failed by slippage of the nails back through their insertion sites at loads less than body weight. Nails with deployable distal fins failed by the fins' cutting out distally through metaphyseal bone at 1.5 times body weight, and nails with proximal and distal interlocking screws failed only at loads of nearly four times body weight. Failure in these cases occurred by fracture at the base of the femoral neck, cutting out of the proximal screw, or bending of the nail within the fracture site. No failure of the distal locking screws was reported.

All of the systems tested demonstrated low rigidity in torsion. Ender nails and open-section interlocking nails reestablished only 3% of the torsional stiffness of the intact femur but with a closed-section nail, the torsional rigidity of the construct increased to about 50% of the intact femur (225a). Although torsional stiffness values for open-section nails are significantly lower than for closed-section nails, the bone–implant combination shows little residual angular displacement after testing. The nail deforms elastically and then springs back, with only a minor slip in the bone. Deployable fins control rotation as well as distal interlocking screws, even in those cases in which deployment of both fins is incomplete. Conversely, Ender nails show a greater degree of slip, thus allowing residual rotational deformity after release of the load.

Biologic Consequences of Intramedullary Surgery

Fracture healing proceeds mainly by the formation of periosteal callus (178,188,191). Cortical reaming and nail insertion injure the medullary vascular system, resulting in avascularity of significant portions of the diaphyseal cortex (48,49 and 50). The implications in open fractures are obvious, and the risk of infection must be carefully weighed against the necessity of reaming. An implant that immobilizes the fracture, however, facilitates revascularization of the fracture site (190).

In studies of the canine tibia, fractures fixed with an intramedullary rod showed higher rates of whole bone and fracture-site blood flow than comparable fractures fixed with a plate, and they remained elevated for a longer period (188). A delay in the maturation of callus was noted with intramedullary nails, but once union was achieved the biomechanical quality of union was similar in the two groups.

Medullary cortical reaming weakens a bone. Clawson et al. (42) recommended removal of no more than 4 mm and cautioned that the cortex should not be reamed to less than half of its original thickness. Pratt et al. (185) studied the effect of reaming on the torsional strength of cadaver femurs; reaming to 12 mm decreased the maximum torque to failure to 63% of that of matched controls, and further reaming to 16 mm decreased this value to 36%, with the largest increment occurring between 14 and 15 mm. They recommended reaming to less than half the bone's diameter at the midshaft.

Molster (152) studied the effect of medullary reaming and nail insertion on intact rat femurs and found that reaming immediately reduced the strength of the femur by 15%, although the degree of cortical reaming was not specified. Femurs, after the introduction of rigid nails, remained weaker than matched controls

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without nails, indicating some degree of biologic impairment from the intervention. Refracture after removal of intramedullary nails has been infrequent and this is partly related to the absence of multiple screw holes, which are stress risers.

The mechanism of osseous necrosis has been studied and appears to be, in part, pressure-mediated. Stürmer (219) made the following observations after extensive laboratory and clinical investigation:

1. Pressure elevation associated with awl, guide rod, and reamers occurred in excess of 1,000 mm Hg, measured both in sheep and humans (Fig. 6-51).

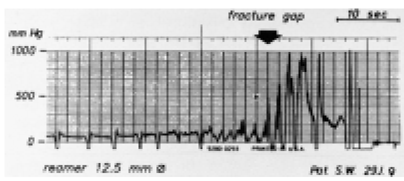

FIGURE 6-51. Intramedullary pressure measured during closed intramedullary nailing of a human tibia. Pressure approaches 1,000 mm Hg in the distal fragment. (From Sturmer K. Measurement of intramedullary pressure in an animal experiment and propositions to reduce the pressure increase. *Injury: AO/ASIF Sci Suppl* 1993;24(3):s7–s21.)

2. High intramedullary pressures caused significantly larger zones of necrosis than a special irrigation-suction reamer, which lowered pressures.
3. High intramedullary pressure was necessary for venous drainage through the periosteal veins; “pulsating intramedullary pressure may be necessary for nourishment of the osteocytes-medullary reaming, and nailing can cause a considerable disturbance of the intracortical transport mechanism.”
4. Irrigation suction system pressure was kept at physiologic levels and thermal effects were reduced; 38.5% of the cortex remained vital, compared with 27.5% in the conventionally reamed group ($p < .05$).
5. Distal venting hole use was ineffective because of the high viscosity of the medullary contents. An unreamed rod raised pressures to 200 mm Hg and embolization occurred.

Thermal effects also contribute to necrosis of bone. Ochsner et al. (164) and others (5,140) have investigated thermal necrosis after reaming and concluded a small canal and a dull reamer are a setup for bone necrosis. The authors recommended reaming with the standard instrumentation (9 mm) only if the medullary cavity has a diameter of at least 8 mm at its narrowest point. Narrow canals should require special reamers or an alternative to reamed intramedullary nailing. Sharp reamers be used. Concern over the harmful effects of reaming has led to a shift of interest to interlocking nailing systems that do not require reaming (90,99,103,129). Rhinelander (190) and others (169,176,188) have shown more rapid revascularization with nails placed without preparatory reaming, compared with nails placed after reaming. In a rat study, Grundnes and co-workers (81) compared reaming and fixation of 1.8-mm canals with 1.6-mm nails and 2-mm nails. Blood flow differences correlating to amount of reaming were noted but were found to be short-lived.

Intramedullary fixation rigidity has been shown to affect fracture healing in animal models. Wang et al. (232) studied the quality and strength of fracture callus in rabbit femurs fixed with rods of varying bending rigidity. Insufficient rigidity produced abundant callus but unreliable and widely variable bone healing. Excessive bending rigidity produced scant callus that demonstrated low-energy absorption when stressed to failure. Intermediate bending rigidity produced optimal callus formation and energy absorption to failure.

Torsional rigidity also influences fracture healing. Molster (153) studied healing of osteotomized rat femurs fixed with intramedullary rods with varying degrees of instability. Union was delayed in the femurs in which rotational instability was the greatest. Woodard et al. (237) reported differences in fracture healing of canine femoral fractures treated with rods of differing torsional rigidity. Closed-section rods were paired with open-section rods of equal bending but differing torsional rigidity, restoring 42% and 12%, respectively, of the torsional rigidity of the femur. Pseudarthrosis occurred in 50% of these femurs treated with the more flexible slotted rods. All femurs treated by closed-section rods were united at 6 months. Grundnes et al. (81) also found torsional flexibility was inhibitory to fracture union.