CHAPTER **36**

ORTHOPEDIC SCREW MECHANICS

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INTRODUCTION

There are seemingly an infinite number of devices coming to market to capitalize on the foot and ankle surgeon's skill as a biomechanical engineer and structural surgeon. The hardware industry has exploded; literally, with new companies being formed almost overnight to sell the latest or greatest devices. Most of these devices are derivatives of existing instrumentation, with claims of advancements or preferential for isolated situations. Arguably no device has seen more iterations than the bone screw. Cannulated, non-cannulated, varied pitch, headless, titanium, self-drilling and tapping, absorbable, multi-piece, and tapered versions are all currently available. These modifications may deliver specific advantages, or disadvantages, as the case may be.

Given the current evolution of the device, these "improvements" warrant a closer look. The science of human anatomic repair must be understood from the surgeon's standpoint, not from the ability to commercialize an apparatus. This paper seeks to explain the mechanics of a screw in its simplest form, beginning with a discussion of the physics of simple machines and extending these principles to development and usage of the screw specifically in foot and ankle surgery, in an effort to better understand the effect of modifications as it relates to real world usage.

PHYSICS

The orthopedic screw (Figure 1) can summarily be defined as a simple machine used by the surgeon to impart force on a bone during the course of skeletal treatment. A machine is a device arranged to change the magnitude, direction, or point of application of moving forces. Motion is an essential part of a machine; without it, we have no machine, but a structure. Therefore a screw is but a structure until put into use. An ideal machine is one in which the parts are considered to be weightless, frictionless, and rigid. Real machines are not ideal, but ideal machines aid thought and analysis, and in many cases are adequate approximations, so they are quite useful. A simple machine is a machine from which no part can be removed without destroying it as a machine. A mechanism is a machine considered solely from the point of view of its motions, kinematically, without consideration of loads. A structure transmits force without motion.

Every machine has an input and an output, and the output is a modification of the input, not a simple replication of it. A machine is a processor or transformer in some sense. The motion of the output is fully constrained by the motion of the input, by their kinematic connection. The force at the input is called the effort, and the force at the output, a load. The mechanical advantage, abbreviated simply as the advantage, is the ratio of the load to the effort. The velocity ratio is the ratio of the movement of the load to the movement of the effort, in linear displacement or rotation. In an ideal machine the product of the advantage and the velocity ratio is unity. There is a trade-off between force and speed. In a real machine the product is less than unity. As a consequence, an ideal machine in equilibrium (when the effort and the load balance) can be moved by the least impetus, as well in one direction as in the other, so the machine is reversible. A real machine, however, requires a certain effort to move it in either direction; it is irreversible, and there is an unavoidable loss of energy whenever it moves.

The inputs and outputs of a machine may be either forces or torques, and a machine may convert one into the other. A torque or moment tends to cause rotation, while a force causes linear motion. The work done is either torque times angle of rotation, or force times distance. The dimensions of torque are force times distance, and this should be carefully distinguished from work, which has the same dimensions. A fundamental property of machines is that the input and output work are the same, except for frictional losses that make the output work smaller. This law of the conservation of energy is a very important generalization.

Simple machines have been classified as lever, wedge, inclined plane, wheel and axle, pulley and screw.

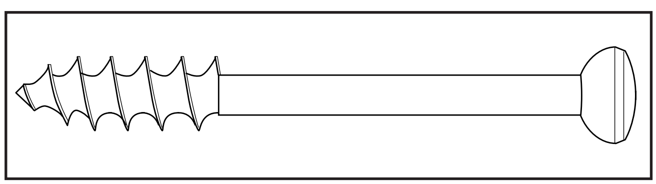

Figure 1. Partially-threaded screw.

Sometimes the wedge and screw are considered special cases of the inclined plane, so there are either four or six simple machines.

Friction is usually the most important reason real machines are different from ideal ones, and in some machines friction plays an essential role. For example, belt drives would not work without friction. Friction cannot be described accurately in a few words, since it is a very complex and variable phenomenon. However, Coulomb's assertions of the general nature of frictional forces between solid surfaces are often adequate, at least qualitatively. The minimum tangential force F that will cause movement between two solid surfaces pressed together is proportional to the normal force N, $F = \mu N$, where μ is the coefficient of friction, which depends on the nature and treatment of the surfaces. The area over which N is spread is of no significance. Once motion starts, the minimum force required to maintain motion is less than F, and may depend on the speed, usually decreasing with increasing speed. This is sliding friction. Rolling friction is much less, practically vanishing if the surfaces in contact are hard and smooth, and there is no deflection. Sliding friction plays significant importance in orthopedic screw insertion as we will see below.

A second limiting factor in machines is the elasticity of their parts. In a rod or bar stressed along its length, the stress, or force F divided by cross-sectional area A, is proportional to the strain, or change of length ΔL divided by the length L. The constant of proportionality Y is Young's modulus. Hence, $\Delta L = FL/YA$. If a machine is scaled up proportionately, the loads F increase as L^3, the areas only as L^2, so the deflection increases as the square of the size. Bending due to transverse forces leads to much larger deflections than simple tension or compression, and these deflections increase more rapidly with size. Because orthopedic screws vary in size, this principle becomes important, especially in the load bearing foot and ankle. The orthopedic screw as a machine is far more likely to be limited by friction and elasticity than by the strength of its parts. It is usually a simple matter to make sure that the parts of a machine are strong enough to support their loads without failure.

THE SIMPLE MACHINES: Inclined Plane, Wedge and Screws

The wedge and the screw can be thought of as special cases of an inclined plane. In the form of the screw, the inclined plane is probably the most commonly used machine. Both have importance in the discussion on bone fixation.

The fundamental concept of an inclined plane is that of coupled, constrained motions in a plane. The reaction to a weight W on an inclined plane that rises a distance b in a horizontal plane a can be resolved into a reaction normal to the plane, N = W cos theta, and a force along the plane, $F = W \sin \theta$. In the absence of friction, W can be held in place or moved by an applied force F. This gives an advantage $W/F = \csc \theta$ over raising the weight vertically. The real inclined plane is modified by friction. If $F < \mu N$, or $\theta < \arctan \mu$, then W will not slide down the plane by itself, but can be pushed down the plane with a relatively small force. The largest angle for which a weight on an inclined plane will not break free and slide down is called the angle of repose. To push W up the plane requires a force of F plus the frictional force, or $F = W(\sin \theta + \mu \cos \theta)$. Hence, the real advantage is usually considerably less than the ideal. If μ is made small (for example, using different metals), the inclined plane is much closer to ideal (Figure 2).

The wedge is an inclined plane where F is applied horizontally to resist a vertical force W by an inclined interface. In the ideal case, with $\theta = 45°$, W = F, and the machine only changes the direction of the force. With a smaller wedge angle, friction will maintain the force W when the input force F is removed. The limiting angle is $\theta = 2 \arctan \mu$, since friction now acts on two surfaces. There are, correspondingly, two varieties of wedge, one for producing motion at right angles, and one for making a firm, but removable, connection (Figure 3).

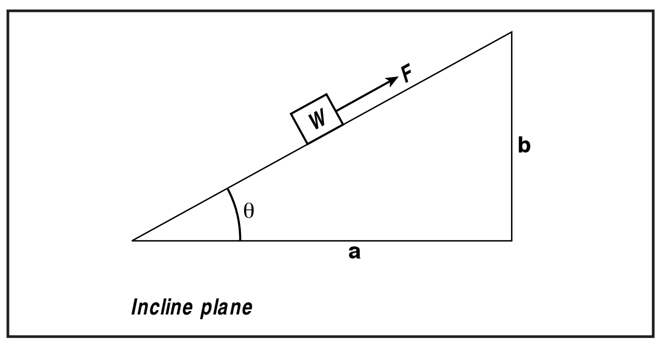

Figure 2. Inclined plane.

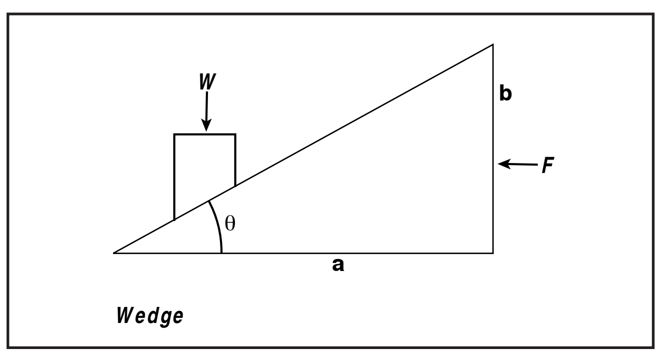

Figure 3. The wedge.

The screw is simply an inclined plane wrapped around a cylinder. The virtues of this arrangement were demonstrated by Archimedes, and the Roman world was the only culture ever to use the screw until modern times. The same two types of applications that we mentioned for the wedge also exist for the screw. By rotating the screw, a bone can be made to move back and forth. As a fastener joining two parts, a nut can be tightened by rotating it, creating a large normal force so that friction will successfully prevent the nut from loosening. If more than one inclined plane is wrapped around the cylinder, the advance in one turn is a multiple of the distance between threads. This gives more threads in the nut, for added strength, or a greater advance per turn, for more rapid motion with the same frictional holding power. For example plastic bottle caps often have multiple screws. They open quickly, but hold securely. Screws are also used to provide a controlled translational motion, as is the case in orthopedic surgery.

THE MATHEMATICAL BASIS FOR THE ORTHOPEDIC SCREW

The principles of statics can be used to analyze the bone screw in detail. It is assumed that the forces are concentrated, although they are actually distributed over the thread surfaces of the screw. They are assumed to act at a radius r, the pitch radius of the screw. The supported load (bone) is denoted by W. θ represents the helix angle. The helix angle is the angle of a screw flight relative to a plan perpendicular to the screw axis. The helix angle references the axis of the cylinder. The advantage of the screw is calculated as $2\pi r/L$, or circumference over pitch L. The pitch of a screw is the distance between two threads (or grooves) from the same point on each thread. On a single thread screw, the lead and pitch are identical. In the diagram "a" represents the coupling of a turning device (Figure 4).

The force F acting at radius r rotates the screw, so the applied torque is $T = Fr$. T may be applied by a force acting on a handle, or by any other method. One rotation of the screw through 2π radians raises the load by a distance L, the lead of the screw. The helix angle of the screw is θ, where $\tan \theta = L/2\pi r$. For a single thread, $1/L$ is the number of threads per unit distance, and L is also the pitch of the screw (Figure 5).

The vector diagram is for the inclined plane unwrapped from the pitch surface. The load is represented by the point P, acted upon by gravity W, the normal force N on the plane, the applied force F, and the frictional force F' (not shown) is directed down the plane, as for impending upward motion of P. L is the lead of the screw. F will be the force required to raise the load W. The forces are resolved in the vertical and horizontal directions. N (normal force) is eliminated, and the result expressed as F/W, or what is equivalent, T/Wr.

The efficiency e is found using the expression for T/Wr, as shown in the diagram. Normally, it may be around 0.15. This means that 0.15 of the input work raises the load, while 0.85 opposes friction. The friction is quite desirable in a lifting screw, since it holds the load steady when the applied force is removed; that is, it prevents overhauling. Roughly speaking, overhauling will not occur when the efficiency is less than 0.50.

ELEMENTS OF A SCREW AND THE EFFECTS ON PERFORMANCE

The world can be described using mathematical formulas. However, for practical purposes we use these equations to better understand what we see, feel, and utilize, but seldom use them in any literal sense in daily life. The preceding discussion serves as the foundation for understanding the intricacies of bone screws. From this platform we will look at the actual device.

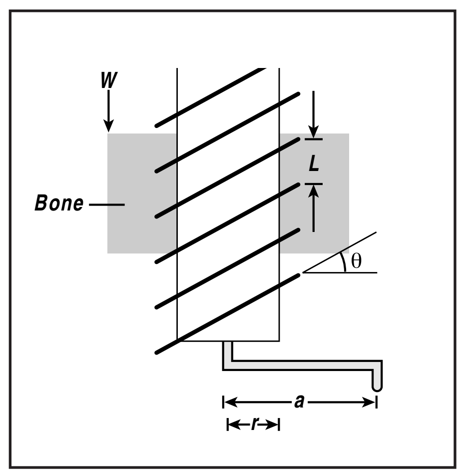

Figure 4. The screw.

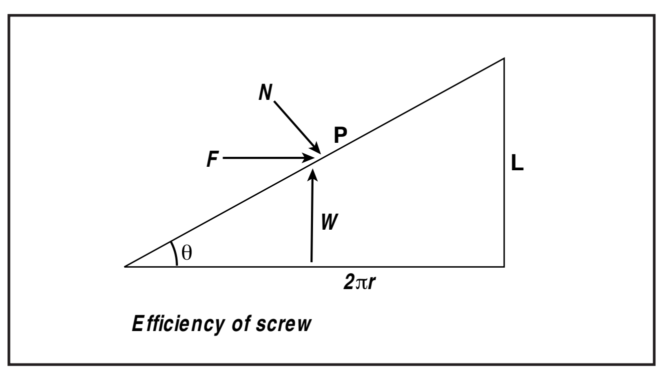

Figure 5. Efficiency of screw.

The mechanical performance of bone screws is determined by their pull-out strength (holding power), compressive force, stripping torque, yield bending moment, ultimate bending moment, and fatigue strength. These parameters are related to the parameters of the screw design, including major thread diameter, minor thread diameter, thread length, pitch, lead and trailing flank angles, helix angle, shaft diameter, cannulation diameter, and material properties (Figure 6).

DIAMETERS

Generally speaking there are two main types of screws, cortical (Figure 7) and cancellous (Figure 8). The distinction rests with the thread pattern and not necessarily with the application; thus at times a cancellous screw may be used in cortical bone, and vice versa. The cortical thread pattern tends to be tighter, that is smaller thread pitch, while screws better suited for cancellous bone have wider or coarser thread pitch. Ideally, the thread to core ratio (ratio of major to minor diameter) is maximized in both situations to increase the interflank contained area.

A mathematical calculation of the volume-contained area (shaded) in Figure 9 could be written as $Ac = 0.7854(D^2 - d^2)L$, where Ac is the interflank contained area, D is the major diameter, d is the minor diameter, and L is average interflank length. Given that the contained volume has a direct influence on the holding strength of the installed and tightened screw thread, this equation shows that major and minor bone screw diameters have an exponential effect on the interflank contained area.

Cannulation, or hollow-shaft modification of a screw (Figure 10), requires an increase in the minor diameter to accommodate for a guide wire while maintaining enough shaft metal to resist unthreading. This profoundly effects the interflank contained area, and is the basis for a measured decrease in pull out strength (1). However, not all clinical reports demonstrate this effect (2, 3). The effect on pull out strength has been shown to be more closely related to major diameter in most studies, with larger diameters consistently outperforming smaller diameters (4).

THREAD FORMS

There are different types of thread patterns used in screw design. These include a square thread, Acme thread, Vee thread, and buttress thread. For orthopedic bone screws, the buttress thread form is used, with leading and trailing flank angles of 45° and 7°. The 7° pressure angle buttress thread is known as the British standard buttress. Other flank angle arrangements exist (for example, a 20° pressure flank and 45° unloaded flank is known as the Benet buttress) however the 45/7 angular differential provides the surgeon with performance advantages in load applied (bearing stress) and decreased radial (shear) stresses (5). This thread with the 45°/7° thread flank angles has demanding characteristics and the very short thread length of engagement results in each pitch of thread being critical (Figure 11)

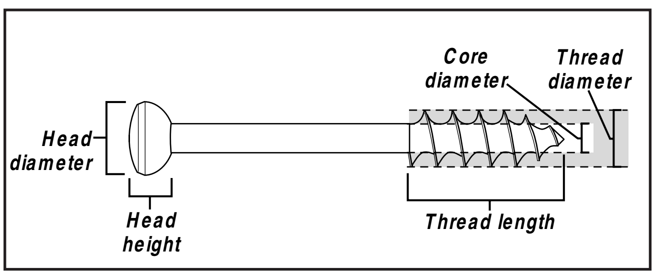

Figure 6. Basic screw.

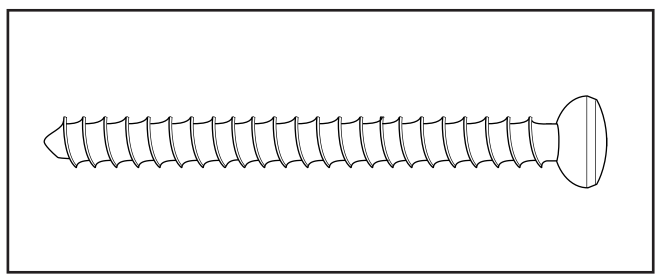

Figure 7. Cortical screw.

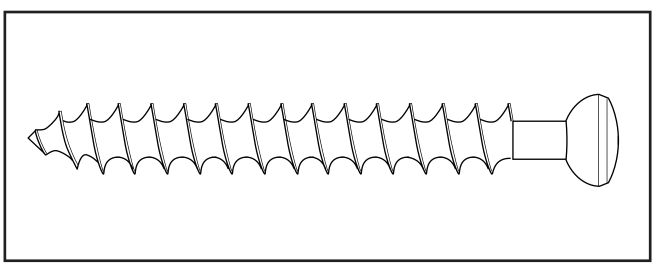

Figure 8. Cancellous screw.

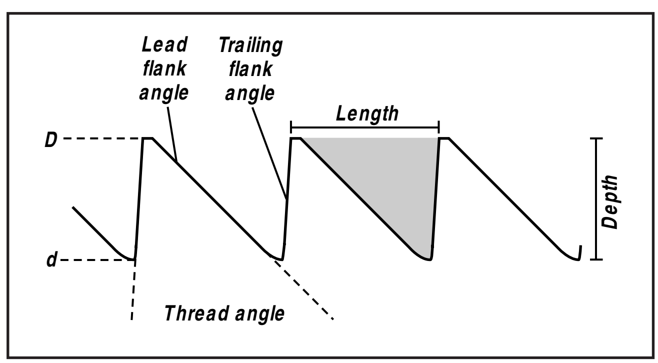

Figure 9. Buttress thread diagram.

NUMBER OF THREADS

Advantage to pull out resistance favors more threads in contact with the bone up to a limit. Beyond six threads there is no appreciable increase in performance for this parameter (6). The number of threads also has no effect on unwinding or "backing out," as noted in the above discussion that the force to over come the coefficient of friction will be the same across all surfaces of the screw.

MECHANICAL ADVANTAGE

There are distinct differences in mechanical advantage between the two types of screws. The mechanical advantage (MA) of a screw can be found by dividing the circumference of the screw by the pitch of the screw, or MA = Circumference/Pitch. In other words, MA is the result of the distance over which effort is applied divided by the distance over which the load is moved. If one thinks of a screw simply as a helical incline plane, the MA is the run over the rise. In actual applications, the screw is often turned by another simple machine such as a lever or a wheel and axle. In this case, the total mechanical advantage is equal to the circumference of the simple machine to which the effort force is applied divided by the pitch of the screw.

For example, suppose a screw with 12 threads per inch is turned by a screwdriver having a handle with a diameter of 1 inch. The mechanical advantage would be calculated as follows: First, the pitch of the screw is determined (Pitch = 1 inch/12 threads = .083); Second, determine the circumference of the handle of the screwdriver (Circumference = 3.14 x diameter = 3.14 x 1 = 3.14 inches); Finally, insert the values obtained into the formula and solve: Mechanical Advantage = Circumference/Pitch = 3.14 inches/0.083 = 37.83.

This equation demonstrates that screws with a smaller pitch (smaller denominator) have greater mechanical advantage. Conversely, a screw with fewer threads per inch will have a lesser mechanical advantage. In the above example, if the number of threads per inch was reduced by half (pitch = 1/6) the MA would reduce to 19.625.

This also demonstrates the effect of the driving instrument on the performance of the machine. There is a clear advantage offered by a larger diameter screw driver, a desirable quality in larger, denser bones. The large amount of rotation delivered at the driver/screw interface with relatively small movements of the driver handle will reduce operator fatigue. This can be advantageous when using screws with tighter pitch, as the smaller lead requires more rotation for each forward advancement of the screw. However, in small or weak bone, when using small screws, or when using screws with impaired ability to stop forward motion (such as a threaded screw head, discussed in more detail below), the added torque delivered by excessively large handles may cause damage to the structure (screw or bone.)

MATERIALS

The two main metals used in the production of orthopedic screws include stainless steel and titanium. It should be stated that there are a variety of proprietary metals that could theoretically be used, however in general surgical stainless steel is 316L grade and titanium is Ti-6Al-4V alloy. Steel has certain mechanical properties that make it suitable for orthopedic screw usage. It is a material that is easily machined for predictable part fabrication. It has a high

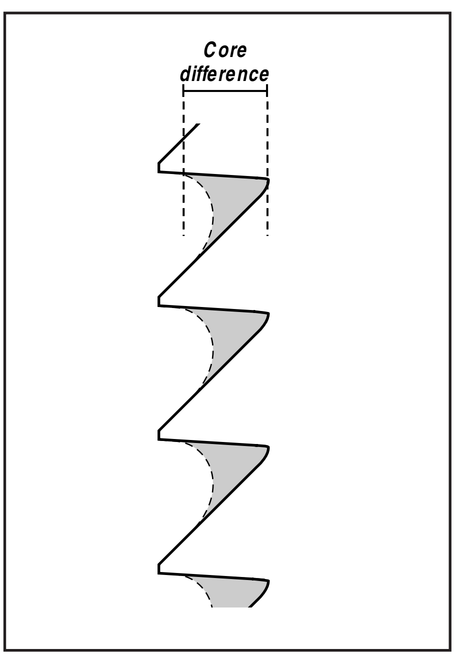

Figure 10. Cannulation.

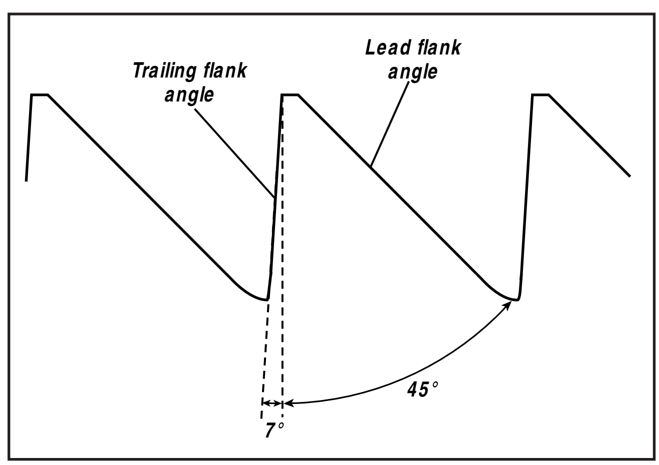

Figure 11. Thread angles.

resistance to pitting and crevice corrosion in chloride and acidic environments. It is also inexpensive. Titanium offers the advantage of improved biocompatibility due to the complete absence of nickel. It is non-ferromagnetic, and has an improved strength to weight ratio (discussed below). Naturally we expect a variance in performance based on the alloy properties.

The static coefficient of friction of each metal on bone differs, with titanium approximately 0.35 and steel 0.42. The higher coefficient translates to increased torque required to overcome frictional stability. This translates to superior implantation characteristics for titanium, while favoring stainless steel in terms of screw back out.

As discussed above, the Young's modulus differs between the metals as well. Numerous alloys of each metal have been used in surgical applications, each with its own material characteristics. In fact, composition analysis is required for ISO (International Organization for Standardization) 9001 certification for medical screw manufacturers, which often can serve as a 'fingerprint' for traceability in case of problems with a material. Alloys are used to enhance certain properties of the metals by introducing impurity atoms to change their internal architecture. It is beyond the scope of this paper to scrutinize all alloys for their surgical usefulness with respect to their material properties. The modulus of elasticity for 316L stainless steel is 193, and 110-114 for Ti-6Al-4V.

Wolff's law states that bones develop a structure most suited to resist the forces acting upon them, adapting the internal architecture and external conformation to the change in external loading conditions. When strain is intensified new bone is formed. When strain intensity is lower than the equilibrium strain deposition activity is less intense than resorption activity and net resorption occurs. Remodeling modifies bone geometric and material properties through a feedback loop. The application of extremely rigid materials will effect the local bone density. The most important principle to understand is that in general a metal with elastic modulus that far exceeds that of bone (GPa 10-20) may impart negative stress-strain characteristics as these materials interact.

Titanium has a lower density compared to steel (4500kg/m3 versus 8000kg/m3). It also has a high yield strength (830 MPa) and ultimate strength (900 MPa) (compared to steel, which has values of 520 and 860 MPa, respectively). When considering these parameters for bone tissue (bone has a yield strength of 104-121 MPa and ultimate strength of 130 MPa) both metals are very useful for stabilizing bone, but titanium has a superior strength to weight ratio. We can also see from this data that there is a variance between yield and failure strengths of the metals. The yield strength is defined as the point at which a material deforms plastically. Prior to the yield point, the material will deform elastically and return to its original shape. Ultimate strength is the maximum stress that a material can withstand before 'necking,' which is often the finality before fracturing. Steel is more ductile, that is it can bend (yield) proportionally more than titanium before fracturing. This property may be advantageous in situations of cyclical loading where the metals undergo repetitive strain application, such as with walking or running.

Absorbable materials have been used extensively in orthopedic screw manufacture, and are generally either polyglycolic acid or poly-L-lactic acid, or a combination of the two. There are also combined enantiomers of polylactate (for example, P(L/D)LA 70/30) to enhance the material properties of PLA and reduce the late adverse reactions at the final stages of polymer degradation. Macrophages and giant cells are considered responsible for the final degradation of polymer debris and contribute to the local tissue reaction seen with these types of implants. The advantage afforded by this type of material is complete absence of ferromagnetism, and they can lessen the need for return surgery to remove hardware (such as in Lisfranc fracture dislocation repair and ankle syndesmosis repair.)

SPECIFIC STRUCTURAL MODIFICATIONS

Pitch Variance

As explained above, there are advantages to using a cortical, or tight, thread pattern over a cancellous, or looser thread pattern. However, the difference between the two patterns can have an interesting effect when arranged along the same shaft in lag-fashion between two structures. The larger lead in the distal segment will generate more screw advancement with each rotation. The smaller lead in the proximal segment will travel a shorter distance with each rotation. The combination then produces advancement at differing rates, which in turn causes a distal structure that the screw is engaging to pull towards the proximal segment of the screw. In surgical application this effect is maximized if the proximal aspect of the screw engages bone of sufficient density, and the gap between bones or segments is less than the difference in pitch times the number of threads in the distal segment (Figure 12).

Head Modification

The head of the screw plays just as important a role as the threads in the development of interfragmentary compression force. As the head engages the proximal cortex, forward momentum of the screw is halted. Prior to this event

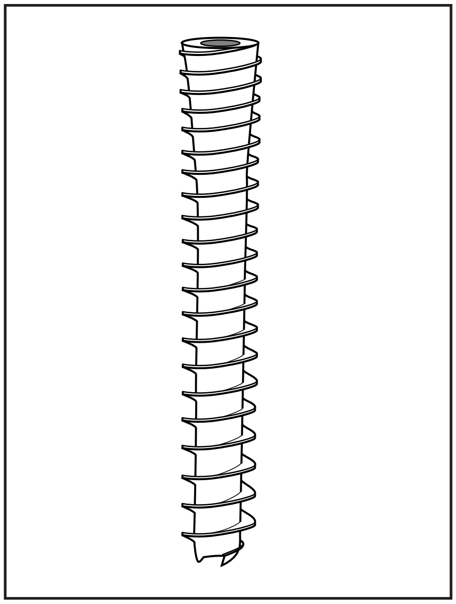

Figure 12. Pitch variance

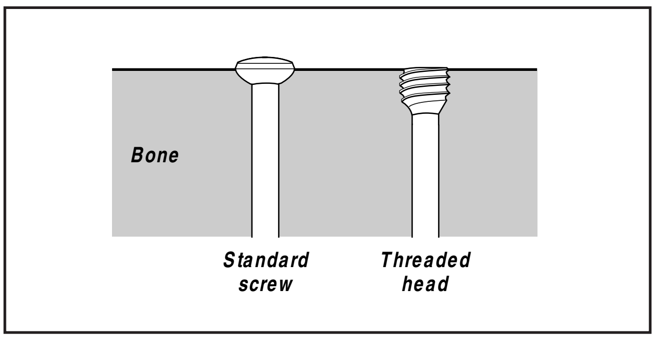

Figure 13. Head modification.

the screw is acting merely as splintage, and the machine functions solely as a wedge. Once the head contacts the cortex, the device then converts to a helical inclined plane as the rotation of the machine continues. At this point one of two things can happen: the threads will cut a pattern into the softer material in which it resides, as a boat propeller chops through water. If enough material is displaced during this process, the normal force acting against the pressure flank is reduced and the screw will "strip" from its hold. The other situation occurs if the proximal segment is released from the screw threads, either through over drilling or by design of the lag screw. When the threads of the screw engage the distal fragment the helical structure will convert rotation to axial movement, causing proximal migration and ultimately compression. If threads are the engine of compression, the head is the transmission, and without it no force can be transferred to the bone.

One common modification is replacing the traditional rounded screw head with a threaded version, usually varied in pitch to the shaft thread, and having a straight or slightly tapered core. This modification is sometimes referred to as the Herbert screw, after T. J. Herbert, a British hand surgeon who invented the screw for use in scaphoid fractures. The intent is to have less screw prominence on the surface of the bone. However, the modification comes at the expense of decreased lag effect, since the forward trajectory of the screw is not diminished as the screw head enters the proximal cortex. This screw is not appropriate for the management of fractures in which compression is of greater importance than the need to avoid prominence of the screw head (7) (Figure 13).

SELF DRILLING AND TAPPING

Self-tapping is the ability of a screw to advance when turned, while creating its own thread. This ability is created sometimes by having a gap in the continuity of the thread on the screw. These edges can cut their own threads as the screw is driven in to the material. Self-drilling screws have a modified tip with a small drill-like flute. They function by having a cutting edge which drills away the material, making a tiny hole for the screw to go into. Generally speaking these types of screws perform exactly as their non-modified counterparts. The advantage provided by a self-tapping pattern is the ability to rapidly insert the screw, such as on a powered driver. Caution should be used however since heat is generated under high speed, which produces a destructive bone debris field around the screw. This can lead to screw loosening since the immediate contact bone material is structurally depleted (8). Self drilling screws are often inserted by hand, and the debris field from drilling has not been shown to affect any functional parameter.

CONCLUSION

In our healthcare system there are several stops on the path from concept based on need to delivery of a technologically advanced product. Technology is not static; it is dynamic, and somewhat amorphous, bending to the needs of its users. Sometimes in the quest to distinguish a device in a crowded field, or when the branch of technology reaches too far from the main trunk, the basics can get lost, and the process becomes one of deterioration. Techniques deteriorate, functionality deteriorates, and outcomes deteriorate. The fundamentals of physics, the science that describes the behaviors of real world objects, should be the starting point in skeletal reconstruction. Understanding the foundation beneath the technology that we utilize is critical to the science and advancement of surgery, because only the surgeon can truly know how a device like a screw and its modifications can and will perform in the real world.

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