# Functional Limitations From Pain Caused by Repetitive Loading on the Skin: A Review and Discussion for Practitioners, With New Data for Limiting Friction Loads

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## ABSTRACT

Reddened painful skin, areas where the outermost layer of skin has been rubbed off (abrasions), and blisters are very common problems observed daily by orthotists, prosthetists, pedorthists, athletic trainers, and a host of other caregivers. These examples of acute skin trauma and the pain that precedes and/or accompanies them are often what determine the limits of an individual's functional performance. When these problems occur within footwear, orthoses, or prosthetic sockets, the cause is repetitive loading—a combination of peak load magnitude and number of loading cycles sufficient to produce significant skin trauma. Walking, running, and many other activities involve movements of skeletal elements relative to surfaces such as shoe insoles, orthoses, and prosthetic sockets. Those relative movements are an inevitable consequence of transferring load through soft tissue and across the skin interface. The loads contain both normal (perpendicular to the skin surface) components and friction (parallel to the skin surface) components. Historically, most instances of skin trauma have been attributed directly to excess pressure and have been dealt with using pressure-management techniques exclusively. Pressure-reduction techniques do often lead to some improvement. However, the assumption that those problems are directly governed by excess pressure is substantially in error, leading to partial solutions and missed opportunities to achieve much higher levels of safe, pain-free function. Research has established the friction component of repetitive loadings to be the major factor governing how many load cycles a given area of skin can tolerate before the onset of pain and acute trauma. To increase comfortable, pain-free function, the magnitude of friction load peaks applied each loading cycle must be reduced. This can be accomplished by reducing peak normal (pressure) load magnitudes and/or reducing the coefficient of friction (COF) operating between the skin and support surface in the problem area. Friction forces over the vast majority of the loaded skin surface area are useful for stability, suspension, and the transmission of functional forces. Friction creates a problem only in very limited areas. This is the rationale for reducing the operative COF in problem areas only. (*J Prosthet Orthot.* 2006;18:93–103.)


**KEY INDEXING TERMS:** friction, blister, repetitive loading, skin trauma, foot wound, diabetes



Repetitive loading occurs in many activities, but most notably it is a characteristic of walking, running, and most other sports activities. It is of obvious importance to prosthetists, orthotists, pedorthists, and, in fact, a wide range of practitioners relating to orthopedics, rehabilitation, and sports medicine. The skin is the interface where load transfers occur. Repetitive loading upon the skin can create "hot spots," blisters, and abrasions, which can cause pain and limit healthy function.

These simple facts have profound implications for caregivers. Knowing how to avoid or delay skin trauma will, in most cases, extend healthy, comfortable function or athletic performance. Repetitive loading causing skin trauma and pain is frequently what defines how far or fast an amputee can walk or run; how much orthopedic support can be provided to the spine, foot, or knee of a growing child; or how far a distance runner can go before acquiring a painful blister. When neuropathy interferes with the protective pain signals that would ordinarily emanate from the dermis, more serious consequences usually result.

There is a long-standing, deeply entrenched concept that excess peak pressure is the governing factor in the generation of skin trauma. However, the research literature clearly indicates that the shear (friction) component of the contact loading is the direct culprit. Pressure is not the direct cause of repetitive loading skin trauma (hot spots, blisters, abrasions, and ulcers). Pressure is a factor that *enables* the friction/shear to reach traumatic levels. The practitioner may ask "what difference does that make?" The difference is that when pressure is thought of as *the* governing factor, a second, equally important factor that enables friction/shear loads to reach destructive, painful levels is ignored: the coefficient of friction (COF). The COF varies widely and depends on the choice of materials interposed between skin and the contacting load.

The almost exclusive focus on pressure as the culprit has led to missed opportunities to enhance and extend function. A thorough, detailed knowledge of the factors and mechanism of repetitive loading skin trauma is basic to understanding how to

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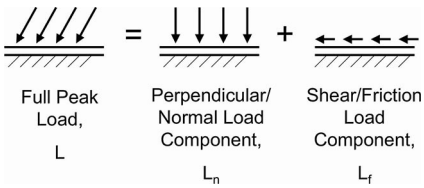


**Figure 1.** During ambulation, the peak load for a given patch of skin consists of a combination of load perpendicular (normal) to the surface and load parallel (tangential) to the surface. Each of those components builds to a peak value at some point in the gait cycle and then retreats. Those peaks are often simultaneous. Peak loads vary greatly from location to location within a shoe or socket.

avoid or delay that trauma and its consequences. The goal of this article is to advance that understanding among caregivers.

During ambulation, each application of load to a given patch of skin consists of some combination of a load component perpendicular to the surface (normal load) and a load component parallel to the surface (Figure 1). Each of these components builds to a peak value at some point in the gait cycle and then retreats. These peaks are often simultaneous but not necessarily so. Loads vary greatly from one point in the gait cycle to another, as well as location to location on the skin.

In everyday practice we might refer to the normal load component as a "pressure" or "compressive" load. The load component parallel to the skin surface is known as a friction, shear, or traction force. In lay vernacular, the words "rub" and "rubbing" are commonly used to describe a load that includes a significant friction/shear component.

Most materials, techniques, and research relating to reduction of skin trauma from repetitive loading have been focused on reducing peak pressure levels. There is certainly a link between repetitive peak pressure levels and skin trauma. The fact of such a linkage is intuitive, and effective measures for ameliorating peak pressure (padding and contouring) have been successfully implemented. However, the true nature of how peak pressure relates to repetitive loading skin trauma has been almost universally ignored for several decades.

Repetitive applications of pure pressure are not injurious unless that loading is extreme. Damaging levels of pressure result in subdermal trauma, with the usual appearance of a bruise or contusion. However, even low to moderate peak load repetitions may become damaging when a significant component of the load is parallel to the skin surface and there is a sufficient number of loading repetitions. When that shear component is large, damage can occur in relatively few cycles. When smaller, a greater number of load cycles is required to accumulate enough trauma to cause pain. The magnitude those friction loads can reach depends on the COF, as well as the peak contact pressure.

In most orthotic-prosthetic practices and academic programs, management of the COF has heretofore not been considered on a par with pressure management as a way to reduce skin trauma and achieve greater comfort, function, and performance. This is an attempt to rectify that oversight by explaining and illustrating how both peak pressure and COF factor into skin trauma.

## REPETITIVE LOADING SKIN TRAUMA RESEARCH

Prosthetic and orthotic professionals have historically considered "pressure areas" to be caused by excessive peak pressures caused by inadequate cushioning or relief contouring. This traditional way of thinking about repetitive loading skin trauma as being the direct result of excessive peak pressures is fundamentally flawed. When reddened skin is observed over a bony prominence, it is more accurate to call it a "friction area." The first definitive, quantitative research on repetitive loading skin trauma factors was published in a pair of articles in 1955.[1,2] British dermatologist Dr. P.F.D. Naylor reported, "There is an inverse relationship between the frictional force and the number of rubs required to produce a blister" (p 342). In other words, the higher the friction force, the quicker a blister is formed. Not surprising, but the full import of this has been largely ignored.

Sulzberger et al.[3] conducted an investigation in the 1960s that corroborated and expanded the scope of Naylor's findings. Naylor and Sulzberger used dozens of human subjects in their investigations during a time when there were few restrictions to prevent consenting adults from being subjected to temporary pain and injury for the sake of science. The skin of the shins and palms of scores of volunteers was rubbed back and forth to the point of pain, blister, and/or open abrasion. In 1972, Akers and Sulzberger[4] published similar results and included an examination of the effects of moisture and temperature on blister formation.

The invention of the resistance strain gauge revolutionized force measurement in the 1950s. That technology allowed the experimental apparatus needed by Naylor and Sulzberger to be much smaller, lighter, and more precise. Both used precision apparatuses to press a pad against the skin with a constant load and move the pad back and forth in a rubbing action. Simultaneously, they measured the friction loads between pad and skin. They deliberately varied those friction loads without changing the pad pressure. They did that by changing skin surface conditions they knew would affect the interface COF. Naylor and Sulzberger consistently found that when they lowered the peak friction loads by applying talc or oil to reduce the COF, it took a greater number of load cycles to produce skin trauma and pain. Conversely, they found that an increase in the COF, such as is caused by moderate amounts of moisture from perspiration, hastened skin trauma and the onset of pain.

Some of Naylor's data indicate that a relatively modest 30% reduction in COF would approximately triple the number of load cycles the skin could withstand before sustaining abrasive trauma. He demonstrated that vastly different rates of skin damage are possible with no change in the normal (compressive) component of the rub load. This finding, in 1955, still has not received due vigorous attention from most orthotic-prosthetic researchers and educators.

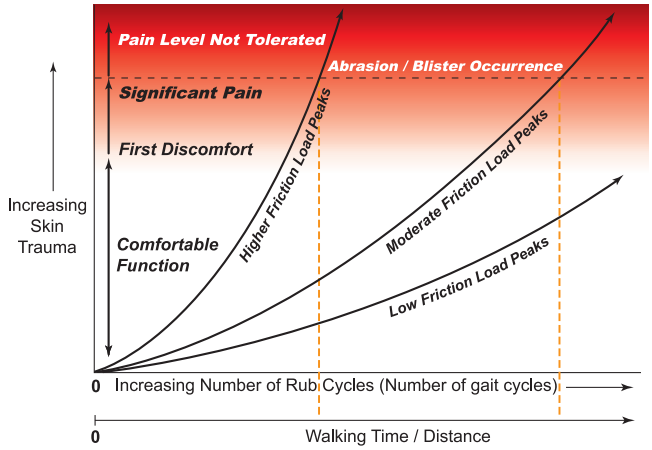


**Figure 2.** Qualitative representation of the relationship between friction load repetitions and discomfort or injury. The vertical axis represents increasing amounts of accumulating skin trauma. The horizontal axis represents the number of repetitive friction load cycles. The three curves illustrate the findings of Naylor and Sulzberger—that load repetitions containing higher levels of peak friction cause skin trauma more rapidly and lead to discomfort in fewer load cycles.

Dr. Naylor did not explicitly say so, but it can be safely assumed that he could have varied the peak friction load by varying the normal (compressive) load on the rubbing pad. That was intuitively clear to him as it is to us. What his research indicates is that peak friction loading is the directly governing factor. Pressure and COF come into the equation because they both determine the peak friction load magnitude.

Figure 2 illustrates how the relationships Naylor discovered play out in real life. As skin trauma increases, discomfort intensifies. Higher friction load peaks lead to pain more quickly (fewer load cycles); lower friction load peaks allow more function (a greater number of load cycles) before the onset of discomfort.

## RUBBING, DISCOMFORT, AND FUNCTION LIMITS

Three short scenarios follow, in which repetitive loading causes skin trauma that limits safe, comfortable function.

### CASE EXAMPLE 1

After a particularly long, tough game of racquetball, one of the players removes his court shoes, exposes his feet to the cooling air, and pays close attention to a very sensitive red area under his right first metatarsal head. He decides to cancel plans to play in a tournament the following day.

### CASE EXAMPLE 2

A man in his early 50s has been living successfully with a diagnosis of diabetes for 20 years. One Saturday he goes to the state fair, makes a very long day of it, and has a great time. When he finally returns home and goes to bed that night, he discovers that one of his socks has a large bloody spot near the toe. A foot inspection reveals a serious wound.

### CASE EXAMPLE 3

A 35-year-old transtibial amputee goes shopping at the megamall with a friend. After four hours, her prosthetic socket becomes uncomfortable. Upon arriving home, she removes her prosthesis, and the hot, sore, red skin over her fibular head begins to cool and feel better. She decides not to use her prosthesis for the rest of the day.

The individuals in these three examples have different diagnoses and activity levels, but they have one thing in common. The rubbing loads and the number of repetitions have been great enough to do skin damage, causing discomfort and placing limits on their function. When protective sensation is not adequate, rubbing can destroy the skin and proceed to deeper, more serious trauma. The location of the accumulating skin damage is commonly called a "pressure area" or "hot spot," an area of reddened, obviously irritated skin within the prosthetic socket, orthosis, or shoe. "Pressure areas" occur on people of all functional levels, from the most marginal ambulators to the most elite athletes.

The racquetball player's skin had absorbed enough repetitive load trauma to put it in the discomfort zone depicted in Figure 2. The same is true for the shopper. The fair-goer proceeded right into serious trauma without knowing it. His damaged peripheral nervous system failed to warn him.

For discussion purposes, assume that the amputee saw her prosthetist about the discomfort in the area of her fibular head, and her prosthetist was able to reduce the peak friction load. After allowing her skin to recover, when she returns to the mall she will be on a slower skin trauma accumulation curve farther to the right on the graph of Figure 2. She may experience no discomfort after 4 hours of walking. If the orthotist or pedorthist caring for the man with diabetes had been able to reduce walking friction peaks under that callused metatarsal head, the fair-goer would have passed into the trauma zone later in his day, or perhaps not at all. The researchers' conclusive statements point unequivocally to friction as the repetitive loading skin trauma culprit.

It is important to point out this does not mean that the normal load component (pressure) is unimportant. It does say practitioners need a more complete understanding of friction, its contributing factors, and how it causes skin trauma to better extend comfortable, safe function for their clients.

## BASIC PRINCIPLES OF FRICTION

Friction is the force that resists the sliding of one surface across another. Imagine that a measured load can be applied to one side of the block shown in Figure 3 resting on a surface. The graphs illustrate that the friction force will exactly match the pull force (in the opposite direction).

However, there is a limit, called the limiting friction load (LFL). If the force on the block exceeds the LFL, the friction force cannot continue to match it, and the block begins to slide. Once sliding begins, the friction force resisting sliding

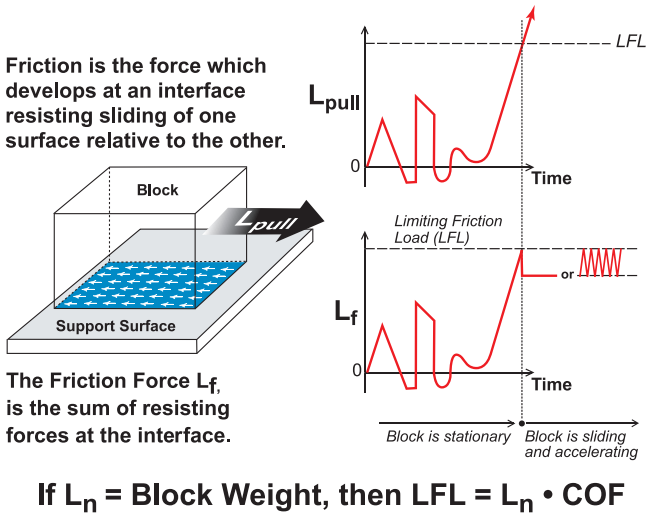


Figure 3. Classic friction experiment, showing a temporal graph of the "pull" load (upper right) and corresponding resisting friction load (lower right). Within limits, the friction force will exactly match the pull force. However, when the pull force on the block surpasses the limiting friction load (LFL), the friction force cannot continue to match it, and the block begins to slide. Once sliding begins, the friction force resisting sliding typically decreases to a lower level. Under certain conditions, the friction load will jump up and down in a series of rapid stick-slip motions.

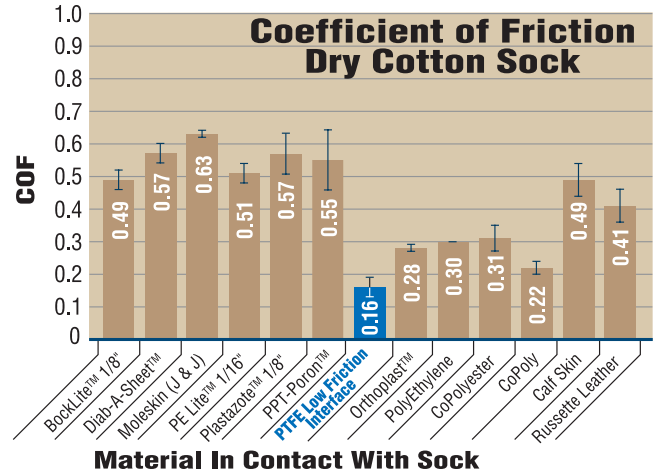


Figure 4. COF measurements for cotton stockinet in forceful contact with each of several materials commonly used in orthotics and prosthetics, under dry conditions. (BockLite, Otto Bock Health Care GmbH, Duderstadt, Germany; Diab-A-Sheet, Langer Biomechanics Group, Inc., Deer Park, NY; Pe-Lite, Fillauer LLC, Chattanooga, TN; Plastazote, Zotefoams PLC Corp., Surrey, United Kingdom; PPT-Poron, Rogers Corp., Rogers, CT; PTFE, polytetrafluoroethylene; Orthoplast, Johnson & Johnson, New Brunswick, NJ; CoPoly, polypropylene with 5–10% polyethylene added.)

typically decreases. (Under certain conditions the sliding may not be smooth. If the system has the right elastic characteristics, movement may be a rapid series of stick-slip actions, as indicated in the graph. This stick-slip motion is the sort of sliding that occurred in Dr. Naylor's experiments.) The LFL is equal to the normal load component ($L_n$) multiplied by the COF. So, because $L_f \leq$ LFL and LFL $= L_n \times$ COF, it follows that $L_f \leq L_n \times$ COF.

Different material combinations can have very different resistances to sliding. Exactly what causes one pair of surfaces to slide easily over one another (without the aid of lubricants) and another pair of materials to greatly resist sliding is not fully understood. We do know that friction characteristics are determined by a combination of physical and chemical surface characteristics.

Although friction may have complex origins, there is a simple, reliable way to experimentally quantify the friction characteristics of a pair of surface materials in contact with each other. Imagine one material fixed to the surface of an inclined plane. The other material covers the bottom of a weight block. The incline of the plane is gradually raised from horizontal. At some angle, the weight block begins to slide. The coefficient of friction for that pair of materials is equal to the trigonometric tangent of the angle of incline where sliding occurs. For example, if that angle is 45°, the COF is 1.0. Initiation of sliding at a 30° angle would mean a COF of 0.58. The trigonometric tangent value automatically gives the ratio of friction force to normal force components at that point when sliding begins.

The COF values for sock-cushion material combinations commonly found in orthotics and prosthetics have not been measured and reported in the past. Because these values are so important, an attempt was made to obtain COF measurements for cotton upon various O&P materials. The weight block mentioned earlier was covered with cotton stockinet and tested in combination with various materials on the inclined plane. The COF values for a cotton sock tested with most of the typical plastic foam materials used in shoes, orthoses, and prostheses are in the range of 0.5 to 0.6 or higher (Figure 4). It is important to note that the data reported in Figure 4 relate to dry conditions. The presence of moisture can profoundly change COF.

It is important to recognize that friction inside a prosthetic socket, shoe, or orthosis is not categorically problematic. It is only a concern in the few areas where red skin, discomfort, or wounds indicate peak friction force levels are high enough to be a problem. In all other areas, friction performs the valuable function of adding stability, control, and/or suspension without causing a problem. So, friction is not something to be avoided everywhere. The greatest benefit is obtained when friction is managed—that is, reduced in certain locations only.

It must also be emphasized that the COF relates to a specific *pair* of materials at an interface. The presence of a third substance at the interface can significantly alter the COF. The interfering substance most commonly and unavoidably encountered within a shoe, orthosis, or prosthetic socket is moisture, which almost always increases the COF, sometimes dramatically (Appendix).

It may be helpful to note that the science of friction should not be confused with lubrication. A lubricant physically in-

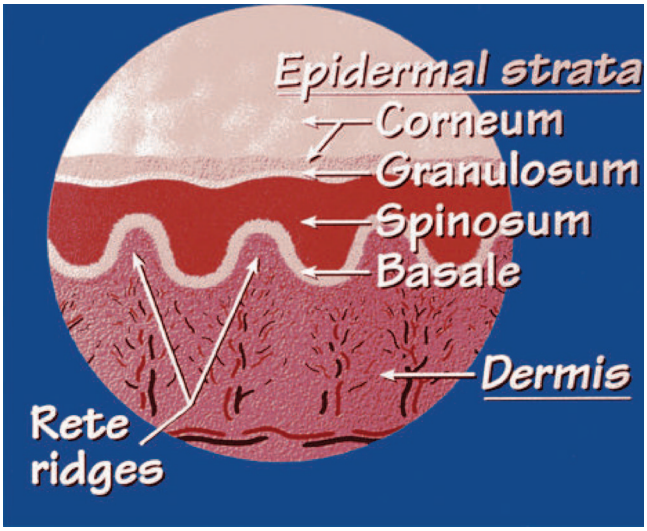


**Figure 5.** Basic anatomy of the epidermis. Cells are produced in the basal cell layer, changing form as they migrate outward, and slough off about 28 days after they were first created.

tervenes between the pair of materials, preventing direct, forceful molecular contact between the two materials.

## EPIDERMAL ANATOMY, REGENERATION, TRAUMA, AND RESPONSE

Abrasions and blisters are not the only significant consequences of repetitive loading on the skin. Spence and Shields[5] noted in 1968 that "[b]listers, callosities and ulcers are primarily caused by excessive shear forces, commonly referred to as friction, acting on the skin" (p 428). To discuss these in context, it is helpful to understand the anatomy and physiology of the epidermis.

The outermost layer of the epidermis is the corneum, beneath which are the granulosum and the spinosum (Figure 5). The innermost layer of the epidermis is the basal cell layer. The convolutions in that layer are called rete ridges. All new epidermal cells are produced in the basal cell layer. Those cells migrate outward after generation, changing form, from spinosum to granulosum and finally to thin, flat corneum cells that flake off about 28 days after they were first created in the basal cell layer.

In addition to corroborating Naylor's findings, Sulzberger and colleagues designed part of their research to learn more about the exact nature of repetitive loading skin damage and exactly which dermal layers were involved. They removed very small, full-thickness skin samples at various stages of repetitive loading trauma. Microscopic examination of those skin samples revealed that trauma first accumulates in the form of micro tears in the stratum spinosum (Figure 6A). As the load repetitions continue, the tears grow in size and begin to coalesce. The process continues, and a cleft is formed (Figure 6B). The cleft is parallel with the skin surface, and the corneum and granulosum layers form a roof over the cleft. Blisters with sturdy roofs can form readily within palmar and plantar skin, and, if the roof is unfractured, the cleft fills with serous fluid (Figure 6C). However, the corneum and granulosum layers of the epidermis on most other parts of the body are thinner and weaker. In those areas, the corneum and granulosum are abraded away at the same time the spinosum cleft is complete.

When the rate of skin damage is low enough or when there are sufficiently long interruptions of the repetitious loading, skin has time to adapt.[6,7] Adaptations include greatly accelerated cell generation by the basal cell layer, which results in epidermal thickening. On the hands and feet, the result is callusing. However, the skin of most other areas of the body reacts differently, becoming thicker but with a fissured roughness.

These adaptations reduce the likelihood of blisters and abrasions. However, a new set of callus-related problems can be created if these adaptations persist too actively. This last is especially true for people with long-standing diabetes and related neuropathic conditions that often include exuberant callus formation on the feet (Figure 7). Callus production can be reduced by reducing friction loading on the callus area. However, cell production by the basal cell layer does not decrease or cease immediately. That delay, coupled with the 4-week cell migration process, means that the benefits of friction reduction under calluses take 30 to 60 days to be fully realized.

## SOFT TISSUE LOADING

Figure 8 qualitatively illustrates an approximation of the sequential sagittal plane motions of the distal anterior por-


**Figure 6.** Repetitive loading trauma tends to focus within the spinosum layer of the epidermis. Early damage involves microscopic tears in the stratum spinosum (A). Continued trauma causes tears to enlarge and combine, forming a cleft parallel to the skin surface (B). The corneum and granulosum layers form an intact roof over the cleft (palmar and plantar skin) or are abraded away (thinner, weaker skin areas) as the process continues. If the roof is unfractured, the cleft fills with serous fluid, forming a blister (C).

Figure 7. Example of a wound on the foot of a person with diabetes. Adaptations of the skin to low levels of trauma include thickening and callusing. Calluses may develop cracks, allowing infectious agents to penetrate to vascular levels. They can also become so thick, hard, and "peaked" that they injure deeper lying tissues (as shown here).

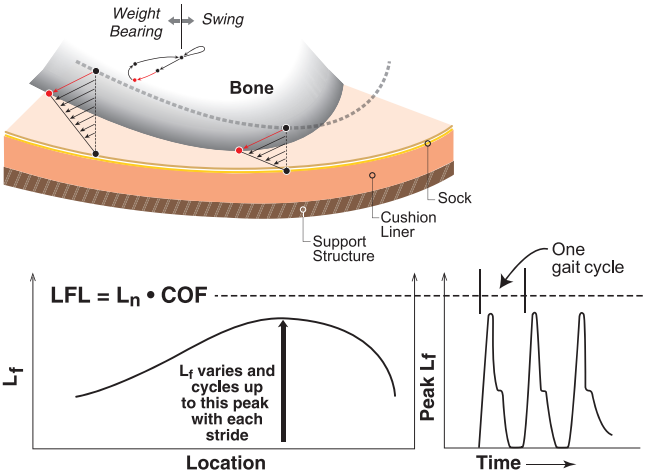


Figure 9. Representation of bone, soft tissue, and support surface within a shoe, indicating bone movement and soft tissue shear through a gait loading cycle. It is assumed that there is no slippage between skin and sock or sock and liner. In other words, the friction load does not reach the limiting friction load.

tion of a tibia inside a prosthetic socket during a typical gait cycle. These images assume no slippage at either the skin-sock or the sock-liner interface. In Figure 8A, imaginary lines are drawn from two locations on the bone through the soft tissue to the skin surface and sock. The arrows in subsequent frames indicate the movement of bone and soft tissue from the unloaded condition to their location at each stage of loading. Figure 8D represents a loading stage at or shortly after mid-stance. There is an angle created by the lines indicating the unloaded versus loaded soft tissue locations. The size of that angle is a reflection of the magnitude of the shear distortion to which the soft tissue is subjected. The shear distortion and friction force are greater where the bone is closer to the surface.

Figure 9 is a general representation of bone, soft tissue, and support surface as might exist within a shoe. The bone could be a portion of the calcaneus or of a metatarsal. The lines and arrows indicate bone and soft tissue movement as the situation progresses from the unloaded condition to a fully loaded stage. Note that both compression and shear occur in the stage of the cycle when loads peak. If it were possible to measure the distribution of the friction loading on the skin along the line of this cross-section, a distribution curve could be depicted. This assumes no sliding occurred between skin and sock or between sock and liner. To the right

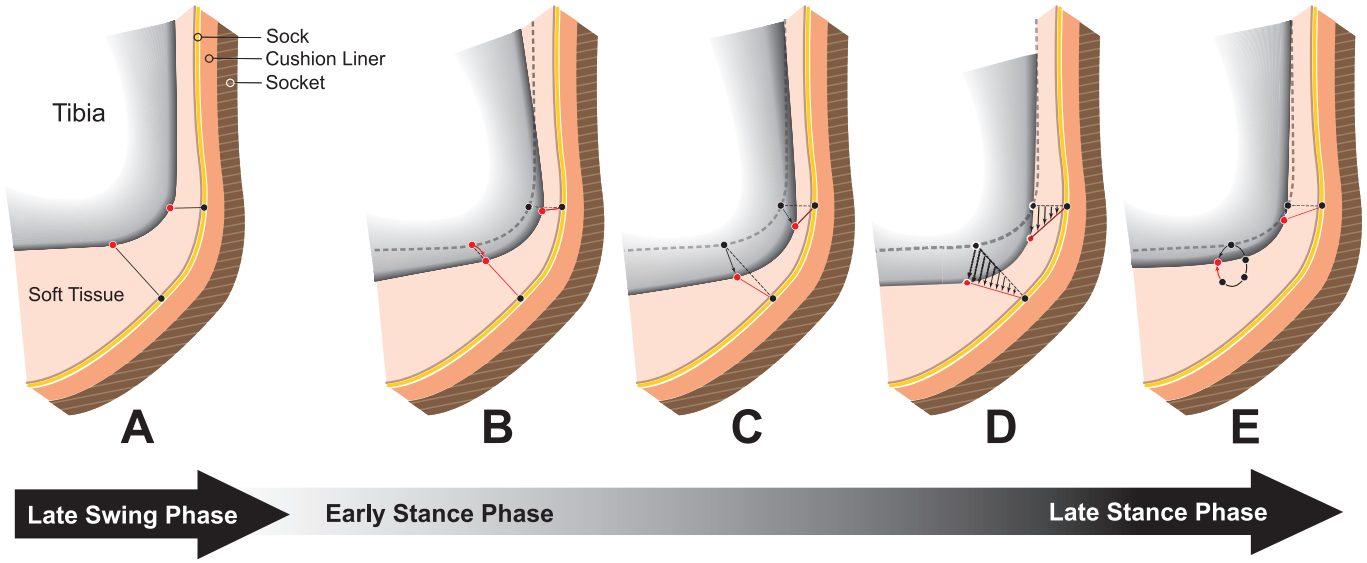


Figure 8. Qualitative illustration of the sequential sagittal plane motions of the distal anterior portion of a tibia inside a prosthetic socket during the gait cycle. The arrows indicate the movement of bone and soft tissue from the unloaded condition to each stage of loading.

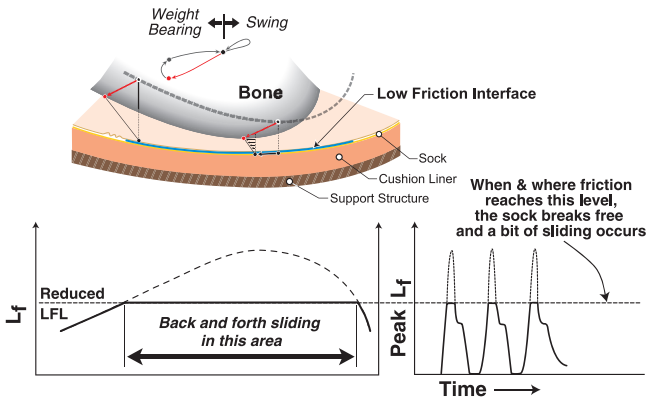


**Figure 10.** When the limiting friction load (LFL) is reduced by application of a low COF patch, the sock begins to slide when the friction/shear load reaches that lower LFL. Less soft tissue shear displacement and lowered friction load peaks result in less skin damage per cycle. Changing the COF does not alter contact/support pressure.

of that graph is an arbitrary temporal representation of how the friction force increases and decreases with each stride. Because there is no sliding, it can be assumed that the LFL is higher than the friction force peaks.

Because the friction load, $L_f$, cannot exceed LFL and LFL = $L_n \times$ COF, consider reducing the COF by about 30% in this area using a low COF patch (Figure 10). At the point in the loading cycle when the friction load in this problem area reaches the lowered LFL, the sock will slide a bit on the low COF interface patch. The reduced COF causes a truncation of the friction force distribution curve (Figure 10, left). Likewise, the tops of the peak friction loads are cut down to the new LFL level (Figure 10, right). Shear strains on skin and other soft tissue are reduced because of that release. Reduced shear displacement and lowered friction load peaks translate into less skin damage per cycle.

If COF intervention is confined to areas known to be potentially at risk, the useful, suspending and stabilizing friction load is preserved in the majority of the socket or shoe. Lower friction load peaks in at-risk areas mean a lower rate of skin trauma accumulation per gait cycle. Increased comfort, function, and performance are the direct result.

Consider the transtibial amputee of Case Example 3. We speculated about moving her to the right on the graph shown in Figure 2 so she could walk farther and longer before experiencing discomfort in the area of her fibular head. Dr. Naylor's data showed that a 30% reduction in COF resulted in the skin being able to take approximately three times as many load cycles before injury occurred. Referring to the COF measurements shown in Figure 4, COF reductions of 50% and more are possible. Naylor's data indicate that a COF reduction of 50% would greatly extend the comfortable function of that amputee at the megamall. This might also prevent the open wound on the foot of the diabetic man in Case Example 2. In time, friction management may even reduce his callus formation because repetitive friction loading stimulates callus formation.

## FRICTION MANAGEMENT TECHNIQUES IN COMMON USE

There are many ways to reduce friction forces between the body and various types of footwear, orthoses, and prostheses that have been used over time. Some interventions directly address the friction coefficient, whereas others are based on the principle of absorbing motion in the structure or mechanism before it reaches the skin.

Creams and oils have been used to lower friction. Unlike the localized application of a low COF interface patch, they tend to spread beyond the area of application, and their beneficial effect on the COF is short lived. Beyond 3 hours, they may actually cause an increase in COF because they tend to increase skin hydration.

Residual limb sheaths have been used to increase comfort in certain situations. The COF between sheath materials such as nylon and common sock materials typically is lower than that of the skin-sock or sock-cushion interfaces. Some distance runners achieve an equivalent effect with a sheer inner sock. When several interfaces are in parallel, the one with the lowest COF determines the limiting friction load. So, the release action described earlier will happen at the sheath-sock interface at a lower LFL. The potential downside of using sheaths is that lowering the COF over the entire surface of the residual limb may sacrifice some needed suspension or rotational stability.

Another approach has been to mechanically isolate the primary socket from some of the twisting and/or pistoning action of the rest of the prosthesis by special mechanical linkage. A design for a suspended inner socket or "double slip socket" was patented in 1885.[8] Tamarack Habilitation Technologies, Inc. (Blaine, MN) has used a similar concept for several people with extreme amputation levels (bilateral hip disarticulation and pelvectomy).[9] The flexible mechanical linkage between the fabric inner socket and the rigid base socket keeps the two in close proximity in a way that allows limited relative movement. Any sliding occurs at low friction levels between the fabric inner socket and the base shell.

Pylon adaptations on the market partially eliminate the transmission of torsional and longitudinal loads between foot and socket. All of these techniques involve absorbing a certain amount of slippage or relative movement that reduces peak friction loading against the skin. Absorbing motion anywhere between the skin and the walking surface represents an extra energy cost, and some loss of precise control.

## SOFT TISSUE MOBILITY AND SKIN TRAUMA

Just as the physical properties of materials used by caregivers vary and contribute to skin trauma, discomfort, and limited function, so do the physical properties of anatomical materials. The most important of these variables is skin "mobility," and it relates to the material science term "shear modulus." A higher mobility corresponds to a lower shear modulus. To appreciate natural variations in skin mobility, compare the mobility of the skin on the back of your open

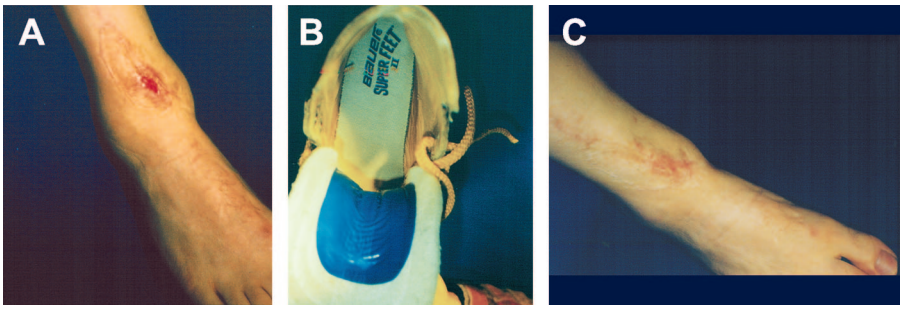


**Figure 11.** Case example of secondary injury to adhered scar tissue caused by friction loading. After complete healing of an open anterior ankle wound, the patient experienced a friction injury to an area of limited-mobility skin (A) while playing hockey. Modification of the skate with a low-friction interface patch inside the tongue (B) lowered the limiting friction load at the anterior ankle while preserving the contact pressure necessary to provide stability. The limiting friction load was lowered enough to protect the skin. Thereafter, he played hockey without any recurrence of the trauma experienced earlier (C).

hand with that of the palm. Compare shin skin mobility to plantar skin.

Skin grafts and adhered scars highlight the importance of mobility. When skin is adhered to underlying anatomical structures, it has limited mobility. In engineering terms, the shear modulus is greatly increased. Any significant shearing action will cause sliding at either the skin-sock interface or the sock-cushion interface. If that sliding occurs at high friction load levels, the skin is rapidly traumatized. We must ensure that when that sliding occurs, the LFL between the sock and contact surface is low enough to protect the skin from high friction forces.

Figure 11 documents a case of secondary injury to adhered scar tissue caused by friction loading. The patient broke his distal tibia in a summertime accident. Complications from excessive swelling inside the patient's cast led to the necessity of a skin graft. Eventually the wound closed. The following winter, during a game of hockey, the subject experienced discomfort in the area of the graft. During the earlier healing process, evidently part of the skin graft covering the injury had adhered to underlying structures. That skin was not mobile and was rubbed off, resulting in an open wound (Figure 11A). The LFL was reduced in the wound area by lowering the COF with a low-friction, polytetrafluoroethylene (PTFE) film interface patch (Figure 11B). The wound healed, and the patient returned to playing hockey with no recurrence of the problem (Figure 11C).

It is not sliding per se that causes the skin damage. High peak friction loads can occur in the absence of any surface sliding. What determines the rate of repetitive loading skin damage is the peak friction load magnitude. That is controlled by the product of the peak normal load multiplied by the COF at the location in question.

## PRESSURE MANAGEMENT BY LOAD SPREADING

Creating a prosthetic socket with sufficient load-spreading inner contours is the first and most important step toward providing comfortable fit and function. The bony features, when properly cradled, provide a prosthesis with excellent orientation, stability, and control. Providing quality "shape matching" involves both art and science. Obviously, comfortable function and performance within a shoe or orthosis also depends on the surface contours in forceful contact with the body. The principles and techniques involved in generating contact surface shapes that efficiently distribute load are extremely important but outside the scope of this article.

When a bony prominence comes into forceful contact with a support surface cushion, the cushion increases the area of contact and thereby spreads the load, reducing the peak load. How effective a given thickness of cushion material is at spreading that load depends on how well the cushion material characteristics and thickness are matched to the anatomical challenge. In most cases there is little anatomical soft tissue covering a problematic bony prominence. Added variables are the magnitude of the skeletal load and the sharpness (radius) of the impinging bone.

Higher skeletal loads and/or bony prominences with smaller radii require firmer cushioning. If the cushion material is either too firm or too soft, limited load spreading will result (Figure 12). Figure 12A depicts what happens when the cushion material is too firm or too thin for the job. If the material is too soft, the material bottoms-out, producing a similar pressure profile. When there is a better match, a broader pressure profile with a reduced peak magnitude is obtained (Figure 12B). Both of these diagrams assume the cushion to be a foam material. Figure 12C depicts what happens when the cushion material is a gel. Most gel materials are not foamed. They are constant volume materials, so they do not simply compress when pressure is applied. There is movement or limited flow of material away from the area of highest pressure toward the areas of lower contact pressure. The result is a cradling bulge that forms around the perimeter, significantly increasing the area of load distribution. Constant volume materials with low shear moduli distribute load efficiently (Bo Klassen, personal communication, Vancouver, Canada, June 1999). The low shear modulus typical of gel also promotes absorption of torsional and longitu-

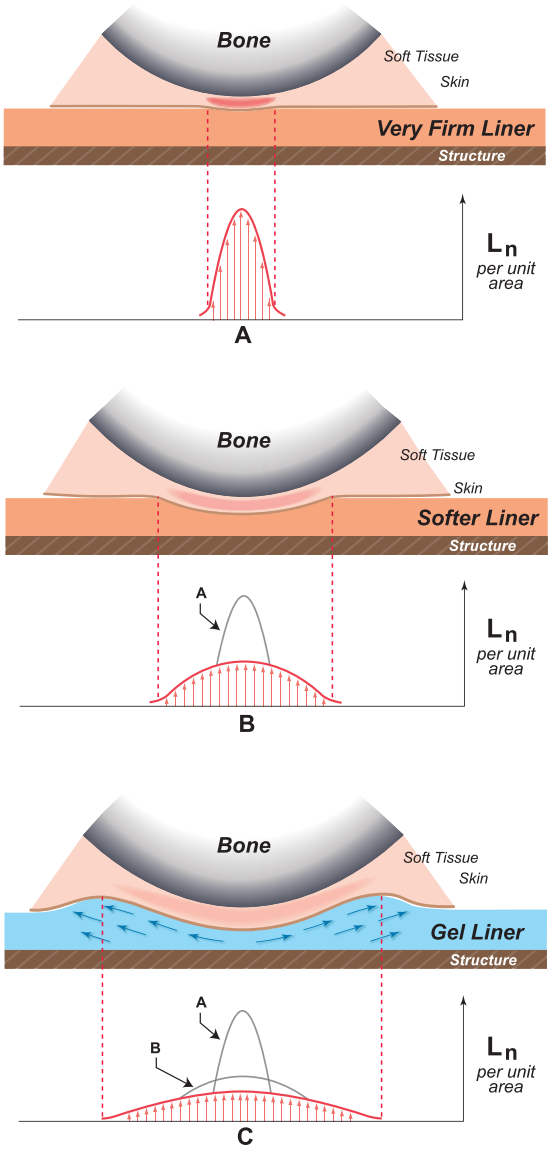


**Figure 12.** Pressure management using cushioning materials to distribute load. If foam cushioning material is too firm, distribution of the load is limited. If it is too soft, the material will prematurely "bottom out" and the load distribution is similar (**A**). Broader load distribution and effective reduction of peak load are obtained with cushion materials properly matching the anatomical need (**B**). Gel materials have constant volume, so they cushion in a more efficient manner. As compression occurs at the apex of a bony prominence, some gel material moves toward the periphery, creating a bulge. This enlarges the supportive contact area, forming a "cradle" under the bony prominence. A very efficient load redistribution is the result (**C**).

dinal strains. The strain absorption acts like an extra layer of adipose tissue and reduces traction or friction forces on the skin. A gel liner is another global response to a local problem.

Both precise contouring and cushioning aim at reducing peak pressure levels. However, it must be emphasized that excellent load transfer contours and firm materials provide excellent, precise control and minimize wasted energy. Cushioning degrades control and energy efficiency. Ideally, cushioning should be used sparingly to augment the pressure management qualities of a carefully shaped surface.

## THE BENNETT ANALYSIS

In his *Transferring Load to Flesh* publications,[10–14] Leon Bennett analyzed compressive and shear stresses that occur in the flesh under and near the edge of a rigid loading surface. He specifically endeavored to model what happens at and near the anterior brim of a transfemoral prosthetic socket. His models analyzed only stresses produced by socket loads applied perpendicular to the skin surface. His models included no bony elements within the soft tissue.

In real life, most prosthetic soft tissue trauma may be found in locations where it is repeatedly loaded between one hard thing (the socket) and another hard thing (a bony prominence within the soft tissue). Also, real-life loading includes components parallel as well as perpendicular to the skin surface. Bennett's analysis attempted to describe how the stress values vary with distance beneath the skin surface and distance from the edge of the loading surface edge. Nearly all of his analyses indicated the highest shear stress levels occurred at the skin surface.

## SUGGESTED RESEARCH

From the time of Naylor's and Sulzberger's research almost a half century ago, surprisingly few practical applications have incorporated their findings. Research into repetitive loading skin trauma and wound genesis must widen its focus to include factors other than peak pressure. With only a few, relatively recent exceptions, researchers typically mention that shear forces are an important factor in skin breakdown but then proceed to measure and relate to peak pressure values and factors.

The most notable of the exceptions is the research done by Joan Sanders and colleagues at the University of Washington, Seattle.[15–18] They have discussed the Naylor and Sulzberger findings in their writings, bringing that information forward to the present. They have designed a shear load sensor and endeavored to measure shear/friction forces at various prosthetic socket wall locations.

Shear sensor technology is still far from miniaturization to the point where it could accurately map shear load distribution. This fact seems to have acted as an almost complete barrier to practical, useful research relating to peak friction loads. However, there are other, very fruitful avenues for investigating the role of friction and the potential value of friction management in reducing ulceration, re-ulceration, and excessive callusing.

Even in the most damaging rubbing environments, localized LFL values may be quite reliably calculated from pressure measurements in conjunction with COF data. The calculation method will not produce an accurate surface map of how those shear forces vary across an entire footbed or socket

because the friction load is equal to $L_n \times$ COF only at locations where friction forces have reached that limit. However, we needn't be concerned with the entire surface. Our concern should be focused on those specific locations where friction forces are high enough to cause wounds, blisters, hot spots, or excessive callusing.

Peak friction loads in at-risk locations can be varied by making known changes in the interface COF at those locations. Such an experiment probably would be preceded by measuring or accessing data regarding the COF for a variety of common cushion and other support materials, each in combination with a variety of common sock materials. Some of this research has already begun.[19,20] From that data, and from conclusions cited here, a number of material combinations could be selected, representing a significant range of COF values (perhaps from 0.16 to near 1.00). Peak pressure values on a pressure map could then be converted to LFL values for the relevant material pairs. These laboratory data and calculations will facilitate the understanding and interpretation of clinical studies.

A variety of useful clinical research hypotheses can be postulated. For example, "Callus production may be significantly reduced by limiting peak friction loads to values below ____," or "The incidence of wound recurrence in diabetic foot wound care may be reduced significantly by managing the coefficient of friction in footwear at the wound site." The work of Naylor and Sulzberger strongly suggests many such hypotheses. Each hypothesis can then be tested by properly designed clinical trials. A population of non-amputee diabetics, for instance, each with a pair of feet, offers an opportunity for arranging a treatment control group. The trials could focus on callus reduction, reducing the incidence of wound genesis, or some aspect of wound healing rate or recurrence prevention. These are all things that have proven to be difficult to manage with current designs and practices. Improvements in health care technology have the potential to make a big difference in the function and health of people in need of this care.

## CONCLUSION

An attempt has been made to "connect the dots" between repetitive loading, pain, and activity limitation. What emerges is a picture of how certain material properties relate to the mechanism and rate of the damage caused by repetitive loading. If we expand our understanding of these concepts, we will be able to develop new designs, new techniques, and new products so affected individuals will be able to perform at higher levels, for longer periods, to greater age, in greater comfort. Practitioners who serve the pediatric population will find that the skin can tolerate higher levels of orthopedic support when friction is properly managed, reducing deformity progression.

There are many health care practitioners who persist in the belief that shear trauma to the skin is something rarely seen—something entirely different from a "pressure area." The extensive and thorough research by Naylor and Sulzberger and others is clear. Friction loading is the cause of repetitive loading damage to the skin. When pressure and friction coefficient are managed, the limiting friction load ($L_n \times$ COF) is reduced. That also reduces the trauma per cycle to at-risk skin, which extends the limits of safe, pain-free function.

## APPENDIX

### THE EFFECT OF MOISTURE ON COEFFICIENT OF FRICTION

The orthotics and prosthetics profession should have COF data available for the various material combinations encountered in clinical practice. The bar graph in Figure 4 of the article is an example of a small amount of such data. It compares the COF of 13 different materials when paired with a cotton sock material. A number of additional weight-bearing surface materials should be objectively tested, each of them interfaced with a variety of sock materials.

The COF measurements reported in Figure 4 were taken under dry conditions. However, conditions inside a shoe, orthosis, or socket are seldom without moisture. Realizing that the presence of moisture will generally change friction characteristics, a limited investigation was performed to determine how cotton sock moisture content affects the COF when interfaced with several common support surface materials.

Figure 13 is a graph of measured COF versus sock moisture content (percent by weight). In this match-up with cotton, two of the support surface materials, PTFE and Plastazote (Zotefoams PLC Corp., Surrey, United Kingdom), had

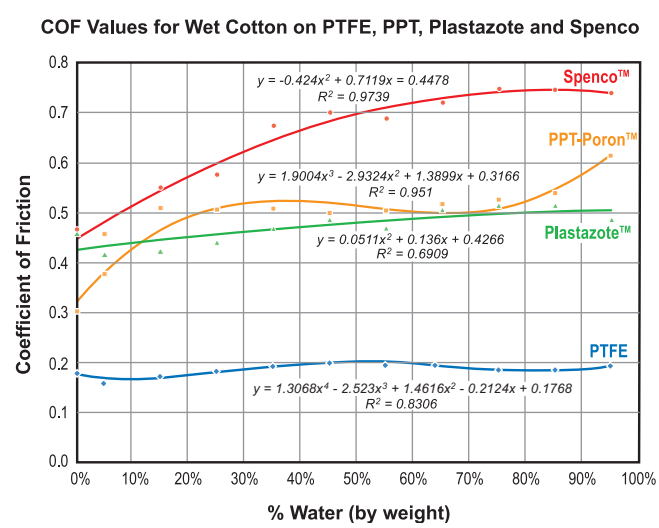


**Figure 13.** Coefficient of friction between cotton and four different support surface materials. Moisture content of the cotton sock is the independent variable and varies from 0% to 100% by weight. The COF of the sock against PTFE and against Plastazote are approximately 0.17 and 0.47 respectively, and those values are not significantly affected by increases in sock moisture content. Spenco an insole material, shows a rather continuous COF increase as the sock gains moisture, and PPT-Poron showed a significant jump in COF to a moisture content of about 35%, then little further increase.

no significant change in their COF values as sock moisture content increased from 0% (dry) to a maximum of 95%. Spenco (Spenco Medical Corporation, Waco, TX) exhibited a rather continuous COF increase with increasing sock moisture content. PPT-Poron (Rogers Corp., Rogers, CT) exhibited a rather sharp COF increase for sock moisture levels to about 35% and little additional increase beyond that.

To understand how much moisture from perspiration exists in an athletic sock, we measured the moisture content of "freshly used" socks from three athletes, acquiring moisture values of 15%, 23%, and 31%. These preliminary tests indicate that COF data would be most valid and useful if measured at a sock moisture content of 25% or 30%.

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