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State-of-the-art research in lower-limb prosthetic biomechanics-socket interface: A review

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Abstract—Scientific studies have been conducted to quantify attributes that may be important in the creation of more functional and comfortable lower-limb prostheses. The prosthesis socket, a human-machine interface, has to be designed properly to achieve satisfactory load transmission, stability, and efficient control for mobility. The biomechanical understanding of the interaction between prosthetic socket and the residual limb is fundamental to such goals. The purpose of this paper is to review the recent research literature on socket biomechanics, including socket pressure measurement, friction-related phenomena and associated properties, computational modeling, and limb tissue responses to external mechanical loads and other physical conditions at the interface. There is no doubt that improved biomechanical understanding has advanced the science of socket fitting. However, the most recent advances in the understanding of stresses experienced at the residual limb have not yet led to enough clinical consensus that could fundamentally alter clinical practice. Efforts should be made to systematically identify the major discrepancies. Further research should be directed to address the critical controversies and the associated technical challenges. Developments should be guided to offer clinicians the quantification and visualization of the interaction between the residual limb and the prosthetic socket. An understanding of comfort and optimal load transfer as patterns of socket interface stress could culminate in socket design expert systems.

Key words: *amputees, biomechanics, computational biomechanics, interface, interface pressure measurement, interface shear measurement, mechanics, prosthetics, socket design, tissue mechanics, tissue responses and adaptation.*

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INTRODUCTION

Surveys have shown that amputees complain about their prosthesis being uncomfortable (1,2). It is not uncommon for amputees to develop skin problems on the residual limb, such as blisters, cysts, edema, skin irritation, and dermatitis (3–5). Discomfort and skin problems are usually attributed to a poor socket fit. Further improvement of prosthetic fitting is required to maximize amputee's comfort and acceptance of the prosthesis.

The socket, as a human-device interface, should be designed properly to achieve satisfactory load transmission, stability, and efficient control for mobility. Some early designs of the prosthetic socket, such as the "plug-fit," took the form of a simple cone shape, with very little rationale for the design. Previous development shows that biomechanical understanding of the interaction between the prosthetic socket and the residual limb is fundamental to the improvement of socket design. With an understanding of the residual limb anatomy and the biomechanical principles involved, more reasonable socket designs, such as the patellar tendon bearing (PTB) transtibial socket, and the quadrilateral transfemoral suction socket were developed following World War II (6,7). These designs intended to provide a more effective distribution of loads around the residual limb. These sockets are so designed that the load-tolerant areas can chiefly take the load, while relief can be given to the sensitive areas. By the 1980s, the so-called hydrostatic weight-bearing principle and the total surface bearing (TSB) concept were introduced. Examples include the silicone suction socket (8) and ICEROSS (9), as well as those incorporating the use of interfacing gel-like materials.

The basic principles for socket design vary from either distributing most of the load over specific load-bearing areas or more uniformly distributing the load over the entire limb. No matter what kind of design, designers are interested in understanding the load-transfer pattern. This will help designers to evaluate the quality of fitting and to enhance their understanding of the underlying biomechanical rationale. Many studies have been conducted to evaluate and quantify the load distribution on the residuum by either clinical measurements or computational modeling.

The skin and the underlying soft tissues of the residual limb are not particularly adapted to the high pressures, shear stress, abrasive relative motions, and the other physical irritations encountered at the prosthetic socket interface. In order to design a good socket fit with optimal mechanical load distributions, it is critical to understand how the residual limb tissues respond to the external loads and other physical phenomena at the interface.

The purpose of this paper is to review the recent studies on prosthetic biomechanics, especially on the socket/residual limb interface, including 1) recent developments in socket pressure measurements, 2) recent investigations on friction-related phenomena and associated properties, such as shear stress, frictional properties of skin, slippage, *et cetera*, 3) computational modeling for residuum tissue stress/strain analysis, and 4) tissue responses to external mechanical loads and other physical conditions at the prosthetic interface.

PRESSURE MEASUREMENTS

The pressure distribution at the interface between the residual limb and the prosthetic socket is a critical consideration in socket design and fit. Pressure measurements within prosthetic sockets have been conducted for about 50 years. The information obtained has been used either to increase the understanding of socket load transfer, to assess the socket design, or to validate the computational modeling.

Interfacial pressure measurements require a proper measurement technique, including the use of transducers, their placement at the prosthetic interface, as well as the associated data acquisition and conditioning approach. An ideal system should be able to continually monitor real interfacial stresses, both pressure and shear, without significant interference to the original interface conditions. A variety of transducers have been developed for socket pressure measurements. They can be classified, based on their operation principle, as fluid-filled sensors (10–12), pneumatic sensors (13–15), diaphragm deflection strain gauge (16–25), cantilever/beam strain gauge (26–28), and printed circuit sheet sensors (29–34), as reviewed by Sanders (35) and Silver-Thorn and colleagues (36).

The techniques for placement of transducers at the residual limb and socket interface can be divided into two categories. They are either inserted between the skin and the liner/socket, or positioned within or through the socket and/or the liner. Only thin sensors, such as the diaphragm deflection strain-gauge sensors (17,18,20,21), the fluid-filled transducers (10), the pneumatic transducers (13,15), and the printed circuit sheet sensors (29–34), are suitable for insertion between the skin and socket. Mounting is relatively easy and it is not necessary to damage the prosthesis. However, for many of these sensors, interference is unavoidable from their protrusions into the socket volume, because of their finite thickness (26,27). The diameter of each sensing element is another important consideration. Too big a sensing element can measure only an average pressure over the area, while too small a sensor may be affected by its edge effects, especially for a stiff sensor. Positioning the transducers within or through the socket with the sensing surface being flush with the skin would make the thickness of the transducer becomes less critical. For such mounting, holes would need to be made on the experimental sockets to recess the transducers (25,26,37–39).

The techniques mentioned above can measure pressures at discrete focal sites because of the size of the sensing cells. Sensor mats with an array of pressure cells make it possible to measure the pressure distribution. However, a piece of material inserted at the interface may change the original conditions. Systems have been commercially designed for *in situ* socket-pressure measurements, such as the Rincoe Socket Fitting System, Tekscan F-Socket Pressure Measurement System, and Novel Pliance 16P System. The F-socket (type 9810 or 9811) transducer is a force-sensing resistor using a mylar substrate for its 0.28-mm-thick strip (40,41). There are 96 individual cells, displayed in an array of 16 rows and 6 columns, covering an area of 155 cm^2. The advantages of this system are its thin and flexible sheet, acceptable sensitivity, resolution, and frequency response (42). The disadvantages usually associated with these sensors are their hysteresis, signal drift, temperature sensitivity, and unknown shear coupling effects (40–42). This system has been used for measuring the pressure distributions at socket interfaces (29,31–34). Houston and colleagues (30) reported a specially designed Tekscan P-Scan transducer with 1,360 cells. Rincoe force sensors are embedded in a polyvenilidyne fluoride strip with a thickness of 0.36 mm (41). This system has a total of 60 cells arranged on 6 separate strips, each comprised of 10 sensors. A report on the use of this system can be found in Shem and colleagues (43). The sensor pad of the Novel Pliance 16P System has 4×4 matrix capacitance sensors with 1-mm thickness. The system allows up to 16 sensor pads to be used simultaneously. There are advantages and disadvantages with each system. The performances (accuracy, hysteresis, signal drift, and the response to curvature) of the above three systems have been compared (40,41).

The pressures reported at the socket interfaces vary widely among sites, individuals, and clinical conditions. For the PTB socket, the maximum peak pressure reportedly could reach about 400 kPa (44), the highest among all the measurements reported. However, the measurements conducted in the last 10 years showed that the maximum interface pressure for PTB sockets during walking was usually below 220 kPa (29,37,38). A wide pressure variation may result from 1) the diversity of the prostheses and fitting techniques used, 2) the difference in residual limb size, soft tissues thickness, and gait style, 3) the different positions studied, and 4) the different characteristics and limitations associated with each specific measurement and mounting method.

SHEAR, FRICTION, AND SLIPPAGE

The biomechanics of the coupling between the skeleton and the socket is an important factor for socket fit. This coupling is affected by the relative slippage between the subject's skin and the prosthetic socket, and the deformation of the residual limb tissues. The tightness of fit could influence the coupling stiffness. Socket shape can change the pressure distribution and the apparent tightness of fit. Generally speaking, a loose fit may allow slippage, which may compromise stability, while a very tight fit may offer a more stable connection, but increase the interface pressures. Another important factor affecting slippage is the friction between the subject's skin and the prosthetic surface. Excessive slippage at the socket interface should be avoided in socket fitting; however, absence of slippage may cause other problems. Amputees might not feel comfortable when a buffer is inserted between the skin and socket to reduce slippage (49). The discomfort apparently resulted not from the pressures, but from the increase in interface temperature and perspiration inside the socket.

Friction is a phenomenon in which tangential force acting between bodies in contact opposes their relative motion or impending motion. Because of the existence of friction, shear forces can be applied to the skin surface. Research related to friction in the prosthetic socket includes 1) investigation of the coefficient of friction of skin with various interface materials (50–52), 2) measurements of shear stresses (37,38,45–48) and slip at the interface (53,54), 3) measurements of the relative motion between the skeleton and the prosthetic socket (55–61), and 4) the contribution of frictional shear to the load transfer.

Frictional properties of human skin have been investigated under various skin conditions (51,62–65), to examine the effects of skin care products (66–68), and to see how friction might affect some friction-dependent manual activities (69,70). Recent studies on skin friction with various interface materials were reported by Sanders and colleagues (51) and Zhang and Mak (52). Sanders and colleagues (51) measured the coefficient of friction of *in vivo* human skin with eight interface materials, using a biaxial force-controlled load applicator. The measurements were conducted on shaved and cleaned skin of the lower limb. The coefficients of friction ranged from 0.48 to 0.89. The coefficients of friction with skin of the eight interface materials are significantly larger than those with sock. Zhang and Mak (52) measured the coefficient of friction of *in vivo* human skin with five materials, namely aluminum, nylon, silicone, cotton sock, and Pelite. The measurements were conducted on untreated skin over six anatomical sites using a Measurement Technology Skin Friction Meter. The average coefficient of friction was 0.46. The value was highest for silicone (0.61) and lowest for nylon (0.37) among the five materials studied.

Measurements of shear stresses at the residual limb/skin interfaces were first reported by Appoldt and colleagues (45). They developed a beam deflection strain-gauge transducer, 11 mm in diameter and 27 mm in length, which could measure the normal force and shear force in one direction. Sanders and coworkers (35,37,46–48) have published a series of papers on the development of their triaxial transducers and their interface stress measurements on transtibial sockets. Shear forces in two directions were measured by mounting metal-foil strain gauges on aluminum beam. The size of the sensing surface was 6.35 mm in diameter but the gross size and weight were quite considerable (37). The transducers have been used to measure the interface stresses on the transtibial sockets to assess the shear stress magnitude (46), the transient shape of the stress waveform during walking (37), and the effects of alignment on these interface stresses (47,48). Williams and colleagues (25) developed a small size (15.9 mm in diameter and 4.9 mm in thickness) triaxial transducer that can measure normal force and shear force in two orthogonal directions. The normal force was sensed by the diaphragm deflection strain gauges. Biaxial shear forces were sensed by magnetoresistors fixed at the center of the disk, which could slide on a cruciform to resolve the shear force into two orthogonal directions. The transducers were further used by Zhang and colleagues (38) to measure the stresses applied on the skin surface at eight locations of five transtibial sockets. A piece of Pelite material was glued on the top of the transducers fixed on the socket wall. A maximum shear stress of 61 kPa was found at the medial tibia area with PTB sockets during walking.

Appoldt and colleagues (53) and Commean and colleagues (54) reported on the measurements of slippage between skin and prosthetic sockets. Appoldt and colleagues (53) developed a slip gage consisting of a pen rigidly held to the transfemoral sockets, whose inking tip was in light contact with the skin. The mark left on the skin was used to assess the slip magnitude and direction. The results indicated that in a well-fitted total-contact suction socket the relative slip was less than 6 mm. Commean and colleagues (54) reported the measurement of slippage between a transtibial residuum and its prosthetic socket using spiral x-ray computed tomography imaging. Lead markers were placed on the socket inner wall and skin surface. The scans were taken under two axial static loading conditions. The results showed that the relative slip increased from 2 mm to 6 mm when the applied load increased from 44.5 N to 178 N.

The skeletal movements relative to the socket are determined by the relative slip between the skin and the prosthetic socket, as well as by the deformation of the residuum soft tissue. Radiography (55–59) and ultrasound (60,61) techniques have been used to investigate the skeletal movements within transtibial sockets (55–57) and transfemoral sockets (58–61). Using the radiographic techniques, measurements were taken under several static load conditions. The results showed that the average movement of the tibia in the proximodistal direction was 22 mm for PTB sockets (57,59) and 11 mm for PTB suction sockets (58). The use of radiographic techniques is limited by the risk of ionizing radiation and often by the static situation. Recently, ultrasound techniques were used to assess femur movements within transfemoral sockets (60,61). This technique allowed recording of such movements during walking using two simultaneously transmitting ultrasound transducers mounted on the socket. The difficulties in performing these measurements routinely in ordinary clinics are that duplicated sockets and prostheses are required, and the need for an expertise-intensive process of ultrasound data analysis (61).

Friction between the residual limb and the prosthetic socket leads to two primary effects. Friction produces shear action on the skin and leads to tissue distortion. Such action may disturb tissue functions and can be harmful to the tissues. On the other hand, the friction-producing shear forces at the skin surface can assist in supporting the ambulatory load and in the suspension of the prosthesis during swing phase. Zhang and colleagues (33) developed an idealized cone-shaped model and a finite element (FE) model using the real limb geometry to predict the effects of friction on the load transfer. Their results showed that the smaller the friction, the smaller the shear stresses, but the larger the normal stresses required to support the same load. Experimental measurements using a Tekscan system apparently confirmed that the pressures measured at a lubricated skin/socket interface were higher than those measured at a normal residual limb/liner interface (33). Hence, reduction of interface friction may not always be a good way to alleviate residual limb tissue problems. An adequate coefficient of friction could be desirable to support loads and prevent undesirable slippage. However, a surface with large friction could experience high local stresses and tissue distortion when donning the limb into the socket, as well as during ambulation. A proper choice of friction would be needed to balance the requirements for effective prosthetic control and minimization of interfacial hazards (33).

COMPUTATIONAL MODELING

Although stresses at the residual limb socket interface can be measured, a full-field experimental evaluation of the load transfer remains difficult. It is anticipated that those difficulties associated with experimental measurements can be overcome by computational modeling, provided an appropriate model can be developed. With the emergence of computer-aided design and computer-aided manufacturing (CAD/CAM) technology, computational modeling is a desirable tool to provide quantitative information on the load transfer between the socket and the residual limb for the purpose of optimal socket design and objective evaluation of the fit. Computational models for socket analysis are mainly based on finite element methods. There are two major advantages in using FE analysis. First, full field information on the stress, strain, and motion anywhere within the modeled objects can be predicted. Second, it is relatively convenient to do parametric analysis for an optimal design.

Since the computational methods were introduced to the prosthetic socket design field in the late 1980s (71,72), several FE models (73–89) have been developed, as reviewed by Zhang and colleagues (90), Silver-Thorn and colleagues (36), and Zachariah and Sanders (91). These models can be grouped into three types (90). The first type involves linear static analysis established under assumptions of linear material properties, infinitesimal deformation and linear boundary condition without considering any interface friction and slip. Models of this type require relatively small CPU time. The second type can be referred to as nonlinear analysis, taking into consideration the nonlinear material properties, large deformation, and nonlinear boundary conditions, including friction/slip contact boundary. Such nonlinear FE analysis normally requires some iterative procedures. While requiring relatively more CPU time, such nonlinear approaches generally yield more accurate solutions. The third type involves dynamic models. Analyses of this type consider not only dynamic loads, but also material inertial effects and time-dependent material properties.

In reviewing the previous FE models, two challenges required to be addressed are 1) modeling of the residual limb soft tissues and 2) the effects of donning procedures with friction/slip interfacial conditions. Biological soft tissues, including residual limb tissues, exhibit complex mechanical properties and may undergo large deformation. The lack of an accurate description of their mechanical properties has limited the development of a precise computational model. The existing data on soft tissue properties were mainly collected using *in vivo* indentation tests (92–99). The material constants under the assumption of linear elasticity, isotropy, and material homogeneity were extracted by curve fitting the indentation force-deformation data with the use of FE technique (75) or by some mathematical formula transfer. The most often used mathematical model is the Hayes' solution (100), based on an elastic analysis of the infinitesimal indentation by a frictionless rigid indentor on an elastic layer bonded to a rigid foundation. The influence of friction between the indentor and the layer surface and the consideration of large deformation were included in a recent study (101). Nonlinear elastic properties, modeled as a Mooney-Rivlin material, have been used for residual tissues in some models (24).

Simulation of donning procedure with friction/slip interfacial conditions is another challenge. In the real situation, the amputee normally puts on the liner first if fitted, and then dons the residual limb into a prosthetic socket. There are difficulties in the simulation of large displacements associated with the donning procedure. To date, most socket rectification is normally simulated by prescribing the displacement boundary conditions at the nodes on the outer surface of the socket or liner (71,75,79,80,82,84,89). Displacement boundary conditions corresponding to the shape of a given socket design are applied to deform the residual limb soft tissue or the liner to conform to the rectified socket shape. There are obvious discrepancies between such simplified simulation and the real donning procedure. Zhang and colleagues (78,79,89) applied interface elements to simulate the friction/slip boundary conditions between skin and liner. Such special four-node elements connecting the skin and the liner by corresponding nodes can be used to simulate friction and slip condition. However, they cannot be used to simulate the donning procedure when there is a large relative sliding between the liner and socket.

Zachariah and Sanders (77) attempted to use an automated contact method, in which correspondence between socket and limb was not required, to simulate the friction/slip interface. Finney (102) attempted to simulate the donning by sliding the deformable residual limb into a rigid socket shell, using a simple idealized geometry.

Further computational modeling for a residual limb and socket system can go in two primary directions. First, computational models with reliable data inputs should be further developed to become more precise, in order to better approximate the real situation. Second, computational modeling can be integrated as part of a clinical system for computer-aided socket design and manufacturing, in order to provide prosthetists with quantitative feedback during virtual socket rectification. Such clinical information must be displayed in a clinically meaningful format and the whole process would need to be user friendly.

A complete prosthesis model should involve not only the residual limb and the socket interface, but also the whole prosthesis. Such a model can be used to discuss the effects of the prosthetic alignment, the foot/ankle joint properties, and the mass distribution of components on the load distribution between the residual limb and the socket, during the loaded support and the socket suspension phases.

TISSUE RESPONSES TO MECHANICAL LOADING

Soft tissues of the residual limb within a prosthetic socket are subjected to a special environment. First, pressures and shear forces are applied by the socket snugly fitted on the residuum, although the limb tissues are not necessarily suited for undertaking such loads. These loads are dynamic and repetitive during locomotion. Second, skin rubbing against the socket edge and interior surface may happen, resulting in intermittent skin deformation and biomechanical irritations. If excessive slip exists between the skin and the socket, tissue abrasion can occur and heat will be generated. Third, residual limb tissues exist often in a high-humidity environment, because the socket intimately fitted on residuum excludes circulating air and traps accumulated sweat. Fourth, the residual limb tissues may suffer from possible chemical and mechanical irritations or allergic reactions to various socket or interface materials (4,103).

Under such an unfavorable and demanding environment, whether the residual limb soft tissues will break down or adapt is a primary consideration in socket design and fitting. If a good skin condition cannot be maintained, the prosthesis can no longer be worn, no matter how accurate the fit of the socket may be. Clinical treatment and socket fit should encourage skin adaptation and avoid breakdown (104). The following review focuses on the response of soft tissues to external loads. The literature reviewed includes studies not only on residual limb tissues, but also other soft tissues that are in contact with external supporting surfaces.

Tissue responses to external forces are complicated, involving tissue deformation, interstitial fluid flow, ischemia, reactive hyperemia, sweat, pain, skin temperature, skin color, *et cetera*. In general, normal physiological forces will not normally disrupt tissue functions. However, an improper application either of an unusually very large force or of a prolonged or repetitive force may damage functions and/or structures. Mechanically, forces applied to skin surface will produce stresses and strain within the skin and the underlying tissues. Those stresses and deformation affect cellular functions and other biophysical processes in the tissues. A very large force may break the skin directly. When moderate static forces are applied to the skin, the underlying blood vessels and lymphatic drainage can be occluded or partially occluded, and oxygen and other nutrients can no longer be delivered at a rate sufficient to satisfy the metabolic requirements of the tissues. Without a sufficient circulation, the breakdown products of metabolism would accumulate within the tissues. If such a condition continues, cellular functions would be compromised and could ultimately fail (105). Tissue breakdown occurs not only on the skin surface, but is often found also in deep tissues (108,109).

Repetitive forces may damage the tissues by accumulating their effects. Although a moderate force may not cause direct and immediate damage to the tissues, repeated applications day after day could initiate an inflammation reaction, and even result in tissue necrosis. When the applied load is within certain windows, tissue adaptation may occur by changing its tissue composition and architecture (106,107).

Besides the load magnitude, other load characteristics, such as direction, distribution, duration (3Ds), and loading rate should be considered in the discussion of soft tissue responses to external loads. The forces applied to the skin surface can be resolved into two components, normal force perpendicular to the skin surface and shear force tangential to skin surface. Some researchers (110,111) suggested that tissue deformation or distortion, rather than mere pressures, are important variables in the study of tissue damage by external loads. When the pressures are evenly distributed over a wide area of the body, damage is apparently less than when loading is applied over a localized area (112). It is generally agreed that an inverse relationship exists between the intensity of the external loads and the duration of load application required to produce ulceration (108,113–115). A number of studies have been presented to theoretically explain such an inverse relationship (116–119). Mak and coworkers (118,119) invoked the physics of interstitial fluid flows induced by a given epidermal pressure to account for the corresponding endurance time. Landsman and colleagues (120) hypothesized that a higher strain rate of tissue deformation may cause a higher pressure buildup in the tissues and a higher elevation of intracellular calcium concentration, leading potentially to more damage to the involved tissues.

Pain

Pain, or discomfort, is the most direct reaction of the human body to excessive external loads. When an abnormally large force is applied to a skin surface, the subject will normally feel some level of pain immediately. Normal sensory function of a human body can often help to avoid a mechanical insult and the subsequent tissue damage. Such sensory feedback can prompt the subject to stop or avoid further application of the loads. Neuropathy can lead to the loss of this function and may result in otherwise preventable damage, such as in the formation of pressure ulcers in diabetic and spinal cord-injured patients.

Load-related thresholds for pain vary with anatomical locations and from person to person. Investigations have been performed to measure the ability of the human body to sustain external forces. The general measurements involve the pressure threshold, i.e., the minimum pressure to induce pain or discomfort, and the pressure tolerance, i.e., the maximum pressure a person can tolerate without excessive effort (121). For residual limbs, the tolerant and sensitive areas have been identified qualitatively (6). Studies have been reported on the load-tolerance levels of the distal ends of residual limbs (122,123).

Microvascular Responses

It is generally believed that ischemia is related to the formation of pressure sores. Ischemia can lead to local malnutrition. Changes in local skin blood supply under various external loading conditions have been studied for a number of years. A series of reports have described the effects of external loads on skin blood flow using radionuclide clearance (124–126), photoplethysmography (127,128), transcutaneous oxygen tension (129–131), and laser Doppler flowmetry (132–139). The results of these studies seem to show that the blood supply would be influenced by the epidermal forces, and the rate and the amount of blood supply would decrease with increased epidermal loads.

Investigations have been done to understand the effects of shear forces in conjunction with normal forces (127,136–138,140). It was noted that cutaneous blood flow was reduced with the increased application of either the normal force or the shear force. The resultant force is a critical parameter in assessing the combined effect of these multi-axial loads (137). Tam and colleagues (138) compared the reactive hyperemia in skin induced by the application of a normal force and that due to the application of both normal and shear forces. It was found that the addition of shear force would increase the tissue recovery time from the effects of hyperemia. This recovery time was taken as indicative of the tissue capacity to accommodate the biomechanical challenges.

Lymphatic Supply and Metabolites

The lymphatic system consists of a complex network of vessels, and presents a drainage route for the transport of excess fluid, protein, and metabolic wastes from the tissue of origin into the circulatory system. External loads may interfere with the normal function of this system. With tissue edema, poor lymphatic function was associated with sore formation (112). Krouskop and colleagues (141) suggested that the smooth muscle of the lymphatics was sensitive to anoxia, and thus the impairment of the lymphatic function combined with changes in the microvascular system could compromise tissue viability through the accumulation of metabolic wastes.

The levels of metabolites in sweat may be used as indicators of the tissue viability status (142,143). Studies showed that epidermal loads could change the amounts and the composition of sweat (144). It was found that there was a significant increase in sweat lactate during loading and a decrease in sweat volume during ischemia.

Skin Temperature

Skin temperature may be taken as a stress indicator for the tissues (145). It was hypothesized that reduced blood perfusion during load application on the skin would be expected to lead to a local fall in skin temperature, and a rise in temperature is expected with the subsequent reactive hyperemia upon load removal. It was suggested that this temperature information might become useful during prosthetic fitting as indicative of the local pressure distribution (146,147). It was shown that tissue temperature decreased as a direct consequence of applied loads (135,145,148,149).

However, whether the temperature can be an indicator of tissue problems is still arguable. Skin temperature is influenced by many factors, which can readily interfere with the absolute surface temperature measurement (145). Schubert and Fagrell (150) found that the temperature increased by 2.7˚C over the gluteus and 1.3˚C over the sacrum when a repetitive normal force was applied to those areas. The contact materials may accumulate heat, which may affect the tissue temperature response. It has been remarked that increasing the temperature by 1˚C can have an effect of increasing the metabolic demands of the cells and oxygen consumption by 10 percent in the associated tissue (151). Skin blood perfusion rate is related to the environmental temperature. Blood flow would increase with a warming environment (152,153).

An increase in temperature may be an early signal for the formation of pressure sores (154). However, apparent controversies still exist. The mean foot temperature of the painful diabetic neuropathic patients is significantly higher than that of the control subjects (155,156). The temperature of the skin at risk of pressure sores was not found to be higher than the other healthy areas (157), although the rate of blood supply markedly differed.

Skin Abrasion

Frictional rubbing is one of the most common insults to which the human skin is exposed (158). It can produce a variety of skin lesions such as calluses, corns, thickening, abrasions, and blisters (159). Repetitive rubbing produces heat, which may cause uncomfortable and detrimental consequence (103). Naylor (158) summarized two kinds of skin reactions to repeated rubbing. One involved skin thickening if the abrasive force is small but rubbing is frequently repeated. The other involved the formation of blisters if the abrasive force is large. It was observed that blisters apparently do not often form on thin skin, but on tough and thick skin (159). Experiments have been conducted to study skin lesions under repetitive pressure with and without the involvement of frictional force (159–161). Results indicated that the addition of friction would accelerate skin damage. Sanders (162) measured the thermal response of skin to cyclic pressure alone and to cyclic pressure plus shear. The results from three normal subjects indicated that the thermal recovery time was apparently higher for combined pressure and shear compared to the values for pressure alone. The apparent additional insults due to shear as demonstrated in this study were consistent with other skin perfusion studies (138).

CLOSING REMARKS

The fundamental goal of prosthetic interface biomechanics research is to achieve optimal and not merely adequate function. Even the most rigorous scientific analyses to date have focused in large part on socket designs based on historical use and proven clinical adequacy. Instrumentation and computer modeling have been useful in illuminating what had only previously been the implied conditions inside prosthetic sockets. However, the most recent advances in the understanding of stresses experienced at the limb/prosthesis interface have not yet fundamentally altered clinical practice (163). Still, it is increasingly necessary for clinicians to cope with new prosthesis designs and materials that do not have the benefit of long histories of successful application. For example, use of new materials such as elastomeric liners and flexible thermoplastic sockets necessarily alter the manner in which load is transferred from the limb to the prosthesis. Improved understanding of prosthetic interface stresses allows us to understand the biomechanical effect of these new interfaces and can help prosthetists to adjust their socket designs to make best use of the properties of new technologies.

For all prosthetic socket designs, the optimal load distribution should be proportional to the ability of the body to sustain such stresses, without crossing the thresholds of pain or skin breakdown. More research is required to obtain sufficient quantitative data to fully document these tissue threshold properties and their dependence on age and pathologies. Without a rigorous understanding of these tissue properties, it would be futile to discuss optimal load distribution and how to achieve that by various prosthetic socket designs.

The CAD/CAM technology for the prosthetic socket may make the socket design and manufacture process more effective and objective. However, the current CAD/CAM systems cannot offer any expert suggestion on how to make an optimal socket design. Further improvement of the systems should incorporate qualification and visualization of the interaction between the residual limb and the prosthetic socket. Computational modeling with further improvements can be a useful tool for this purpose. If the research can accumulate enough information on the relationship between quantified values and the comfort of the prosthesis, CAD/CAM systems can be further developed into expert systems that propose an optimal socket configuration.

The most radical of new prosthetic developments is certainly direct skeletal attachment of limb prostheses through osseointegrated implants. This method completely obviates the need for the prosthetic socket through percutaneous titanium fixtures that transfer load from the prosthesis directly to the skeletal bone. While it may seem that osseointegration renders moot any discussion of prosthetic interfaces, even this radical advance in the state of the art only changes the location and type of the interface problem. New challenges arise from the metal/bone and metal/skin interfaces. The latter juncture is of particular importance because it must artificially provide the critical skin barrier to the environment. Responses of soft tissues to the abnormal stresses at the point of attachment are somewhat related to interface mechanics studied previously.

Prosthetic biomechanics is one of the most challenging areas in the field of biomechanics. There is no doubt that improved biomechanical understanding has advanced the science of socket fitting. However, the most recent advances in the understanding of stresses experienced at the residual limb have not yet led to enough clinical consensus that could fundamentally alter clinical practice. Efforts should be made to systematically identify the major discrepancies. Further research should be directed to address the critical controversies and the associated technical challenges. To these ends, we hope this review article could offer some contribution.

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