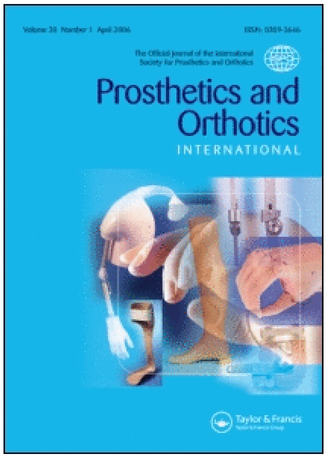

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The thermal conductivity of prosthetic sockets and liners

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Abstract
Elevated stump skin temperatures and the accompanying thermal discomfort are side effects of prosthesis use that may reduce amputee quality of life, particularly in hot or humid surroundings. Lower skin temperatures might be achieved through more effective heat transfer in the prosthesis, a process governed in part by the thermal conductivity of the sock, liner, and socket layers. To assess the thermodynamic properties of currently available components, an instrument capable of measuring the heat flux across a regulated temperature differential was developed. Experimental results show that the thermal conductivity ranged from 0.085–0.266 W/m · °K for liner materials and from 0.148–0.150 W/m · °K for socket materials. The results of this study demonstrate that the prescription of typical multi-layer prostheses constructed with the higher thermal conductivity materials might reduce temperature-related discomfort in patients.

Keywords: *Amputees, thermal conductivity, sockets, liners*

Introduction

A primary function of the lower limb prosthetic socket system is to provide structural coupling between the stump and prosthetic limb during posture and movement. No less important is minimizing the discomfort of the patient. While the modern socket system does provide adequate biomechanical performance, it can also act as an insulating barrier, limiting heat transfer. The resulting elevated stump skin temperatures may lead to discomfort, excess perspiration, and promote stump skin injuries.

There is some survey evidence that suggests thermal discomfort is responsible for a moderate to significant reduction in the quality of life of many amputees (Hagberg and Branemark 2001). Unfortunately, little is known about stump skin temperatures inside prosthetic socket systems or differences between them. It has been shown that donning causes a moderate temperature increase, walking causes a significant increase, and the rest periods following activities must be substantially long to return the limb to resting state temperatures (Peery et al. 2005; Klute et al. 2006). The elevated skin temperatures may also trigger a local thermoregulatory response producing perspiration. In intact individuals, the perspiration

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response is primarily dependent on body core temperature and, to a lesser degree, skin temperature (Wyss et al. 1974; Nadel et al. 1971). However, because amputees have less surface area to allow heat loss, body core and skin temperatures may be more tightly coupled. Regardless of the trigger mechanism, many amputees complain that excessive local perspiration negatively affects the interface between the limb and the socket or suspension system (Legro et al. 1999). Further, the resulting warm and moist conditions inside a prosthesis may be a contributing factor in causing infections (Kohler et al. 1989) and are the same as those found to promote friction blisters in laboratory experiments (Naylor 1955a, 1955b; Akers and Sulzberger 1972).

The solution to this problem is challenging. To cool the human body, several mechanisms can be employed, including convection, radiation, evaporation, and conduction. Unfortunately for the amputee, each of these mechanisms is compromised. Convective cooling of the stump using the circulatory system may be limited by the vascular disease of some amputees, and the natural convection by air is blocked by the prosthesis. Whole body convective heat transfer is also compromised as there is simply less surface area available due to the loss of a limb. Garments and socket systems act as a physical barrier by blocking radiative heat transfer. Evaporative mechanisms are severely limited by the low moisture permeability of most modern socket systems (Hachisuka et al. 2001). Conductive heat transfer through the component layers of the prosthesis is thought to be limited due to an assumption of low thermal conductivity, however, little is known about the thermodynamic properties of the current generation of prosthetic components.

The aim of this study was to measure the thermal conductivity of a wide selection of liner and socket materials. This metric is an important thermodynamic property as it is the material constant that quantifies the amount of heat that can pass through a known volume in a unit of time for a unit difference in temperature through physical contact. The project required the development of an instrument that can measure the heat flow through a test specimen of known thickness while maintaining a regulated temperature differential. This instrument was then used to test a variety of prosthetic component materials used in clinical practice.

Methods

The task of measuring the thermal conductivity of prosthetic liners and socket materials required the design, fabrication, and calibration of a custom-built instrument (Figure 1). The instrument measures heat flow from a hot surface, through a material of interest, to a cold surface. As the thermal conductivity of many materials can vary across a range of temperatures, it is important to measure thermal properties across a temperature differential of clinical interest. The hot surface was maintained at 40°C using five resistive polyimide heaters (Minco, Minneapolis, MN). One of these was used as a central heating element and four were arranged as co-planar guard heaters. To minimize lateral heat losses, four additional heaters were located around the edges of the square samples. To provide a temperature differential across the specimen, a cold surface opposite to the hot surface was maintained at 30°C using a tenth heater and a heat sink. A fan was attached to the heat sink to pump heat away from the cold surface and into the environment in order to create a steady-state heat flow through the specimen.

To ensure a uniform temperature differential, five temperature-sensing thermistors ($\pm$0.1°C, GE Thermometrics, Edison, NJ) were placed on both the hot and cold surfaces (ten total thermistors). These sensors were used to provide feedback signals for heater operation using a custom proportional-integral-derivative control algorithm implemented with LabView (National Instruments, Austin, TX) and five solid state relays (Crydom, San Diego, CA).

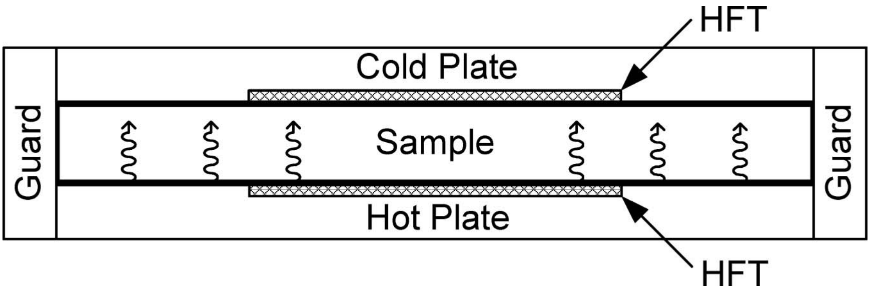

Figure 1. Instrument to measure the thermal conductivity of prosthetic liners and sockets with integral heat flux transducers (HFT).

The outputs from the instrument included the heat flux across the sample measured using heat flux transducers (International Thermal Instrument Company, Del Mar, CA) at both the cold and hot surfaces. A linear potentiometer (Omega Engineering, Stamford, CT) was used to measure the thickness of the sample. Data from the sensors were sampled at 0.6 Hz over a 20-min period for each specimen.

The thermal conductivity (k) of the test specimen was calculated using Fourier's law for steady-state conduction through a specimen of constant area:

$$k = \frac{qL}{A\Delta T} \tag{1}$$

where q is the heat flux across the specimen (mean from the hot and cold surface transducers), L is the thickness of the specimen, A is the specimen area, and ΔT is temperature differential between the hot and cold surfaces.

The instrument was calibrated using Styrofoam^{TM} sheet from the National Institute of Standards and Technology (Gaithersburg, MD) with a known thermal conductivity ($k = 0.034$ W/m · °K) and Teflon^{TM} sheet (virgin electrical grade product #8545K18, McMaster Carr, Los Angeles, CA) with a published thermal conductivity ($k = 0.230$ W/m · °K) (Grigull and Sandner 1984). Data from measurements on these materials were used to determine the heat flow transducer gain. The resolution of the device was calculated at 0.001^{TM}W/m · °K or 1% of the measurement, whichever is larger.

Liner test specimens were prepared by cutting a section of uniform thickness to a square dimension of 100 mm by 100 mm. Thermoplastic socket specimens were fabricated by heating a standard bell (Bulldog Tools, Lewisburg, OH) in a thermoform oven to at least 170°C and then stretching it over a flat surface. Carbon fibre socket specimens were fabricated using epoxy acrylic resin, braided carbon fibre fabric purchased in sleeve form having 4.7 mm wide braids and nyglass, which is 70% nylon and 30% fibreglass, in a typical lay-up (carbon fibre, nyglass, carbon fibre, nyglass). Dibenzoyl peroxide catalyst was added to the epoxy acrylic resin to form a ratio of catalyst to resin of 2% by weight and cured in less than two hours.

To prevent trapped air from influencing the measurement outcome, each test specimen was coated with thermal grease (Thermalcote, AAVID Thermalloy, Concord, NH) and allowed 2 h in the instrument to achieve thermal steady-state. For liners with fabric covers, the influence of trapped air on the measured thermal conductivity may be significant. To explore this influence, specimens of both the OWW Alpha Original and the ESP Aegis Streamline (with fabric cover) were tested with and without thermal grease.

Results

The thermal conductivity of the liner materials ranged from 0.085 – 0.266 W/m · °K (Table I). The Pelite closed cell foam had a thermal conductivity of 0.085 W/m · °K and the Bocklite closed cell foam had a thermal conductivity of 0.091 W/m · °K. The materials used to produce elastomer liners can vary greatly between manufacturers and between product lines, and few details are known about their chemical make-up or the manufacturing process. Liners made from mineral oil gel (manufacturer's description) had thermal conductivities ranging from 0.114 – 0.184 W/m · °K. Liners made from silicone ranged from 0.181 – 0.266 W/m · °K.

The thermal conductivity of the two socket materials, carbon fibre and thermoplastic, were very similar at 0.148 and 0.150 W/m · °K, respectively (Table I).

Both the OWW Alpha Original (3 mm thickness) and the ESP Aegis Streamline (6 mm thickness) were tested to explore the effect of thermal grease on liners with fabric covers.

Table I. Thermal conductivity of prosthetic liner and socket materials.

	Fabric cover	Thermal conductivity (W/m · °K)	Thickness (mm)	Product description
Liner				
Pelite[1]	N	0.085	4.2	Closed cell foam
Syncor, Durasleeve[2]	Y	0.085	3.5	Closed cell foam
Bocklite[3]	N	0.091	6.0	Closed cell foam
OWW, Alpha Original[2]	Y	0.114	3.0	Mineral oil gel
OWW, Alpha Max[2]	Y	0.128	6.0	Mineral oil gel
OWW, P-pod[2]	Y	0.143	3.0	Mineral oil gel
OWW, Alpha Spirit[2]	Y	0.155	6.0	Mineral oil gel
ALPS, EZLiner HP Fabric[2]	Y	0.164	6.0	Silver in gel
Centri, Cushion Liner[1]	N	0.164	3.0	Thermoplastic elastomer
Euro International, Contex-Gel Streifeneder[4]	Y	0.166	6.0	Polymer gel
Freedom Innovations, Evolution SP[5]	N	0.181	3.0	Platinum cured silicone
ALPS, VIVA Sleeve[2]	Y	0.182	6.0	Gel
Medipro, RELAX[2]	Y	0.182	6.0	Silicone with Umbrellan®
Silipos, Explorer Gel Liner[2]	Y	0.184	6.0	Mineral oil gel
ESP, Aegis Streamline[2]	Y	0.187	6.0	Pure silicone
ESP, Aegis Streamline[2]	N	0.189	6.0	Pure silicone
Medipro Sensitive[2]	Y	0.194	6.0	Silicone
ALPS, VIVA Sleeve HP Fabric[2]	Y	0.202	6.0	Gel
Ossur, IceRoss Dermo Seal-in[6]	Y	0.205	6.0	Dermogel®
Euro International, Silicone First Class Liner[4]	Y	0.212	6.0	Silicone
ESP, Aegis Ultimate[2]	Y	0.225	6.0	Pure silicone
Ottobock, Silicone Liner[3]	Y	0.228	3.0	Silicone gel
Ossur, IceRoss Comfort Plus Sensil Gel[6]	Y	0.266	6.0	Soft Sensil® silicone gel
Socket				
Carbon fibre lay-up		0.148	4.2	
Thermoplastic		0.150	4.6	

OWW, Ohio Willow Wood; ESP, Engineered Silicone Products. Suppliers: [1]Fillauer, Chattanooga, TN; [2]Southern Prosthetic Supply, Paso Robles, CA; [3]Otto Bock, Minneapolis, MN; [4]Euro International, Tampa, FL; [5]Freedom Innovations, Irvine, CA; [6]Ossur, Aliso Viejo, CA.

The thermal conductivity of the OWW Alpha Original with thermal grease was 0.114 W/m · °K while without it was 0.083 W/m · °K. The thermal conductivity of the ESP Aegis Streamline with thermal grease was 0.187 W/m · °K while without it was 0.114 W/m · °K.

Discussion

This study reports the measured thermal conductivity of a variety of prosthetic liners and sockets. This property is typically unknown or unpublished for most prosthetic limb components. Accurate measurement of thermal conductivity required the development of a suitable measurement device.

An important consideration when interpreting these results is that a comparison between products with the objective of predicting stump skin temperature requires an assumption of a constant quantity of heat supplied by the stump. Some individuals may respond to a change in prosthetic components by altering their stump cutaneous circulation through sympathetic neural control, thus maintaining the same stump skin temperature in spite of a change in prosthesis thermal conductivity. Certainly healthy subjects have the capacity to vasoconstrict in response to cold environments and vasodilate in response to warm environments. However, there is evidence that diabetes may compromise the thermoregulatory response (Charkoudian 2003). Diabetic patients had a higher incidence for heat illness during heat waves (Schuman 1972; Semenza et al. 1999) and this may be linked to cutaneous vasodilator dysfunction, with (Arora et al. 1998; Veves et al. 1998) or without the presence of neuropathy (Caballero et al. 1999). There is also indirect evidence of impaired thermoregulation from several studies that have found impaired sympathetic neural control of sweating and blood pressure in diabetic subjects (Hilstead and Low 1997; Low et al. 1975; Fealey et al. 1989; Benarroch and Low 1991). These studies suggest that diabetic dysvascular amputees may have a compromised thermoregulatory response and the assumption of a constant quantity of heat supplied by the stump may be a reasonable first approximation for resting state conditions. Traumatic amputees may also have a compromised thermoregulatory response depending on the nature and extent of the trauma.

The prediction of possible differences in stump skin temperatures between different prosthesis requires a model that can accommodate a composite structure. The test results reported here were measured for a single layer of material, whereas the typical prosthesis assembly is a composite structure comprised of multiple layers. The temperature differential (ΔT) between the stump skin and the environment for such a layered structure can be predicted from:

$$\Delta T = \frac{q \cdot \sum\left(\frac{L_i}{k_i}\right)}{A} \tag{2}$$

where the index i refers to the properties and geometry of the component layer. Using this equation, the data reported here can be used to explore the effect different prosthetic components might have on stump skin temperatures if a constant heat source is assumed (i.e., no thermoregulatory response) and steady resting state conditions.

Consider a stump skin temperature of 32.6°C for a subject wearing a carbon fibre socket and a 3 mm thick Pelite liner at rest in a laboratory environment of 23.0°C (Klute et al. 2006). Assuming a constant heat source, changing the carbon fibre socket to a thermoplastic socket would be expected to have little to no effect on stump skin temperature because the thermal conductivities of the two materials are nearly equal (Table I). However, if the patient's liner was changed from a 3 mm Pelite to a 3 mm Alpha Original, the predicted stump skin temperature would decrease from 32.6°C to 31.2°C due to the higher thermal conductivity of the Alpha Original liner. When the patient's liner was switched to a 3 mm Explorer, the predicted stump skin temperature decreased even more, to 29.7°C. Alternatively, when the thickness of the hypothetical patient's Pelite liner was increased from 3 to 4 mm, the predicted skin temperature would increase to 34.4°C. These predictions, based on a simple thermodynamic model, suggest that the choice of socket material has little effect on stump skin temperature, but the liner material and its thickness may produce measurable differences.

While a change in liner material and thickness might produce a measurable change in the stump skin temperature, would such a difference be clinically significant by producing a decrease in discomfort and incidence of injury?

The perception of thermal comfort has been anecdotally reported (Legro et al. 1999) and occasionally measured (Hagberg and Branemark 2001) using survey instruments; however, little is known about the temperature difference necessary to perceive a difference in comfort as a result of different prostheses. For a person acclimated to a particular temperature, research has explored the temperature differences necessary to perceive a rapidly changing thermal stimulus applied to a peripheral limb. Davis studied threshold detection in upper limb amputees in a laboratory environment of (~23°C) (Hunter et al. 2005; Davis 2006). Participants were asked to identify a warming sensation on bare skin with no prosthesis in response to a thermal stimulus that began at a baseline temperature of 32°C and was increased at 0.5°C/s. The subjects detected a warming sensation at 35.7°C; an increase of 3.7°C. In comparison, intact subjects starting at the same baseline temperature but subjected to a slower ramp rate (0.2°C/s) in a somewhat warmer environment (27°C), detected a warming sensation on their forearm at 33.6°C; an increase of only 1.6°C (Divert 2001). The apparently large difference in threshold level might be due amputation, but could also be the result of different stimulus rates or environmental temperatures.

These two studies provide no insight into identifying what stump skin temperature would be perceived as comfortable by an amputee, but can provide some insight into an amputee's perceived state of comfort in response to an activity (e.g., walking) that would generate additional heat and elevate skin temperatures. Additional results reported by Divert (2001) confirm an inverse relationship between the starting temperature of a thermal stimulus and the temperature difference required to perceive warmth. At cooler stimulus starting temperatures (e.g., 30°C), a larger increase in temperature is necessary (~2.5°C) to detect a warming sensation but at a warmer stimulus starting temperature (e.g., 34°C), a smaller increase in temperature is necessary (~1.0°C) to detect a warming sensation. If this same relationship holds for amputees, it suggests that a prosthesis that enables a cooler stump skin temperature to begin with would require a much greater increase in temperature before the amputee would notice a difference. If, for example, an activity such as walking resulted in an increase of 1.5°C (Klute et al. 2006), then the subject whose prosthesis resulted in a 30°C stump skin temperature prior to walking would not notice the difference caused by walking and might be more comfortable than if the subject was wearing a prosthesis with a 34°C stump skin temperature prior to walking. For the warmer starting condition of 34°C, if walking resulted in an increase of 1.5°C, then the subject probably would notice a difference and might claim to be uncomfortably warm. This effect could be achieved by prostheses with different thermal conductivities.

The difference that a change in liner material and thickness might have on incidence of injury is also difficult to predict. *In vivo* research has shown that slightly moist skin generates more friction than either dry skin or very wet skin (Akers and Sulzberger 1972; Naylor 1955b), and there is an inverse relationship between the amount of friction and the number of rubs required to form a blister (Naylor 1955a). For amputees with slightly moist skin inside their prosthesis from local perspiration, the frictional load will be higher and the threshold number of steps to form a blister will be lower. If skin temperatures become elevated, the situation gets worse as there is evidence to suggest that friction blisters form somewhat more quickly when the skin is warm rather than cool (unpublished study by Cortese et al. and reported in Akers and Sulzberger [1972]). Related anecdotal evidence has shown that some amputees have a compromised ability to control skin irritations during periods of warm weather where stump skin temperatures are likely higher than in moderate or cold weather (Legro et al. 1999), supporting the hypothesis that elevated skin temperatures increase the incidence of stump injuries.

A possible limitation of the methods used in this study involved the application of thermal grease to all test specimens to prevent trapped air from influencing the measurement outcome. The thermal conductivity of the OWW Alpha Original and the ESP Aegis Streamline, both of which have fabric covers, was lower when tested with thermal grease than without. These measurements suggest that the thermal grease permeated the fabric covers and displaced any pockets of air that might have decreased the measured thermal conductivity of the specimen. Indeed, the thermal conductivity of the ESP Aegis Streamline with thermal grease applied to the fabric was nearly equal (within 1%) to the thermal conductivity of the ESP Aegis Streamline without a fabric cover (Table I). While application of thermal grease is standard procedure for thermal conductivity testing, liners that can trap a layer of air between the elastomer and the socket may have lower thermal conductivities than reported here. One difficulty to testing fabric-covered liners without thermal grease is that the volume of trapped air is likely to be related to applied load. During gait, the applied load varies and might periodically alter the volume of trapped air and hence the effective thermal conductivity.

The results presented here may facilitate the design of future epidemiology studies exploring the temperature-related effects on amputee quality of life. Components with higher thermal conductivities are expected to result in lower stump skin temperatures, affecting both discomfort and incidence of injury. Further work is necessary to expand the methods reported here to measure the thermal conductivity of prosthetic sock materials. Trapped air in these materials may significantly influence the measured thermal conductivity where the quantity of air may be influenced by load.

Conclusion

Elevated stump skin temperatures and associated thermal discomfort may reduce amputee quality of life, particularly in hot or humid surroundings. The results presented here suggest that some prosthetic components can act as a barrier to conductive heat transfer due to low thermal conductivity. When the clinician assembles a layered structure for patients with temperature-related discomfort, he or she may choose to select components that provide more effective heat transfer. The consideration of heat transfer properties in next generation prosthetic systems might result in the development of more comfortable prostheses and perhaps a lower incidence of skin injuries.

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References

Akers WA, Sulzberger MB. 1972. The friction blister. Milit Med 137:1 – 7.

Arora S, Smakowski P, Frykberg RG, Simeone LR, Freeman R, Logerfo FW, Veves A. 1998. Differences in foot and forearm skin microcirculation in diabetic patients with and without neuropathy. Diabetes Care 21:1339 – 1344.

Benarroch EE, Low PA. 1991. The acetylcholine-induced flare response in evaluation of small fiber dysfunction. Ann Neurol 29:590 – 595.

Caballero AE, Arora S, Saouaf R, Lim SC, Smakowski P, Park JY, King GL, Logerfo FW, Horton ES, Veves A. 1999. Microvascular and macrovascular reactivity is reduced in subjects at risk for type 2 diabetes. Diabetes 48:1856 – 1862.

Charkoudian N. 2003. Skin blood flow in adult human thermoregulation: how it works, when it does not, and why. Mayo Clin Proc 78:603 – 612.

Davis KD. 2006. Personal communication confirming the experiments reported in Hunter et al. 2005 were conducted in a laboratory environment of ~23°C.

Divert VE. 2001. Body thermal state influence on local skin thermosensitivity. Int J Circumpolar Health 60:305 – 311.

Fealey RD, Low PA, Thomas JE. 1989. Thermoregulatory sweating abnormalities in diabetes mellitus. Mayo Clin Proc 64:617 – 628.

Grigull U, Sandner H. 1984. Heat conduction. Berlin: Springer-Verlag.

Hachisuka K, Matsushima Y, Ohmine S, Shitama H, Shinkoda K. 2001. Moisture permeability of the total surface bearing prosthetic socket with a silicone liner: Is it superior to the patella-tendon bearing prosthetic socket? J Uoeh 23:225 – 232.

Hagberg K, Branemark R. 2001. Consequences of non-vascular trans-femoral amputation: a survey of quality of life, prosthetic use and problems. Prosthet Orthot Int 25:186 – 194.

Hilstead J, Low PA. 1997. Diabetic autonomic neuropathy. In: Low PA, editor. Clinical autonomic disorders: Evaluation and management. 2nd ed., Philadelphia, PA: Lippincott-Raven Publishers.

Hunter JP, Katz J, Davis KD. 2005. Dissociation of phantom limb phenomena from stump tactile spatial acuity and sensory thresholds. Brain 128:308 – 320.

Klute GK, Berge JS, Huff E, Ledoux WR. 2006. The effect of rest and exercise on residual limb skin temperatures. American Academy of Orthotists and Prosthetists Annual Meeting and Scientific Symposium. Chicago, IL: AAOP.

Kohler P, Lindh L, Bjorklind A. 1989. Bacteria on stumps of amputees and the effect of antiseptics. Prosthet Orthot Int 13:149 – 151.

Legro MW, Reiber G, Del Aguila M, Ajax MJ, Boone DA, Larsen JA, Smith DG, Sangeorzan B. 1999. Issues of importance reported by persons with lower limb amputations and prostheses. J Rehabil Res Dev 36:155 – 163.

Low PA, Walsh JC, Huang CY, Mcleod JG. 1975. The sympathetic nervous system in diabetic neuropathy. A clinical and pathological study. Brain 98:341 – 356.

Nadel ER, Bullard RW, Stolwijk JA. 1971. Importance of skin temperature in the regulation of sweating. J Appl Physiol 31:80 – 87.

Naylor PFD. 1955a. Experimental friction blisters. British J Dermatol 67:327 – 342.

Naylor PFD. 1955b. The skin surface and friction. British J Dermatol 67:241 – 248.

Peery JT, Ledoux WR, Klute GK. 2005. Residual limb skin temperature in transtibial sockets. J Rehabil Res Dev 42:147 – 154.

Schuman SH. 1972. Patterns of urban heat-wave deaths and implications for prevention: Data from New York and St. Louis during July, 1966. Environ Res 5:59 – 75.

Semenza JC, McCullough JE, Flanders WD, McGeehin MA, Lumpkin JR. 1999. Excess hospital admissions during the July 1995 heat wave in Chicago. Am J Prev Med 16:269 – 277.

Veves A, Akbari CM, Primavera J, Donaghue VM, Zacharoulis D, Chrzan JS, Degirolami U, Logerfo FW, Freeman R. 1998. Endothelial dysfunction and the expression of endothelial nitric oxide synthetase in diabetic neuropathy, vascular disease, and foot ulceration. Diabetes 47:457 – 463.

Wyss CR, Brengelmann GL, Johnson JM, Rowell LB, Niederberger M. 1974. Control of skin blood flow, sweating, and heart rate: role of skin vs. core temperature. J Appl Physiol 36:726 – 733.