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Clinical measurement of normal and shear stresses on a trans-tibial stump: characteristics of wave-form shapes during walking

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Abstract
Stresses on the surface of a stump within a prosthetic socket during walking can potentially traumatise stump tissues. To gain insight into stresses and design parameters that affect them, normal and shear interface stresses were measured on three unilateral trans-tibial amputee subjects during walking trials. During stance phase repeated characteristics in wave-form shapes from different subjects were apparent. They included "loading delays", "high frequency events (HFE's)", "first peaks", "valleys", "second peaks", and "push-off". Characteristics did not necessarily occur at the same time from one step to the next but their timings matched well with events in shank force and moment data which were collected simultaneously. For "plantarflexion" and "dorsiflexion" alignment changes, the above wave-form characteristics were still pesent but their timings within the stance phase changed. The physical meaning and relevance of the characteristics to stump tissue mechanics are discussed.

Introduction
Two types of localised stresses are generated between a stump and a prosthetic socket during ambulation: (i) *normal stresses* are perpendicular to the interface, and (ii) *shear stresses* are in the plane of the interface. Normal and shear stresses are important because they can traumatise stump skin tissues. Excessive static normal stress has been shown to cause blood flow occlusion (Daly *et al.*, 1976) which can result in necrosis and ulceration (Levy, 1962). Trauma from dynamic shear is apparent as separation at the dermal-epidermal junction, followed by fluid deposition and blister formation (Stoughton, 1957; Hunter *et al.*, 1974).

It is the authors' hypothesis that it is not exclusively the magnitude of the stress that is important in causing tissue breakdown. Instead, it is proposed that it is a combination of parameters. For example magnitude, frequency, and loading in other directions simultaneously may possibly all contribute to breakdown. The basis for this hypothesis comes from tissue mechanics literature and clinical experience. Naylor (1955) found that it took less work (defined as $\int$ Force · Number of cycles) to induce friction blisters on the anterior tibial surface of normal subjects if a high number of cycles at low stress were applied compared to a low number of cycles applied at high stress. Lanir and Fung (1974) demonstrated that under an applied uniaxial strain, stress was higher if displacement in the perpendicular direction was restricted. From clinical experience on trans-tibial amputees, Radcliffe and Foort (1961) noted that: excessive pistoning of the stump in the socket resulted in an abrasion; excessive friction irritated existing bursae; adventitious bursae formed over bony prominences and around tendons; and under excessive load bursae become distended, tender, and infected.

The purpose of this paper is to report characteristics of interface stress wave-form shapes from unilateral trans-tibial amputee subjects ambulating with prosthetic limbs. Both normal and shear stresses are described. Effects of those characteristics on stump tissue mechanics are discussed. Results for altered alignment settings and differences between static and

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Table 1. Physical descriptions of amputee subject stumps.

PARAMETER	SUBJECT #1	SUBJECT #2	SUBJECT #3
Dimensions:			
Length (from patellar tendon to distal end):	14.6cm	15.2cm	11.4cm
Antero-posterior diameter at patellar tendon:	7.5cm	7.6cm	8.6cm
Medio-lateral diameter at femoral condyles:	9.2cm	9.5cm	10.2cm
Unusual characteristics:	very bony stump; little soft tissue, especially distally; very dry skin antero-distally	prominent medial distal osteophyte	excessive superficial tissue; prominent peroneal nerve; 1.3cm deep clefts in suture line scars at antero-distal end
Scar locations:	single scar on lateral surface	antero-distal and lateral surfaces	three parallel axially-oriented scars on antero-lateral and antero-medial surfaces; also a single scar on the distal end
Adherent scar tissue sites:	none	antero-distal	distal
Common clinical problems:	ingrown hair antero-medially; folliculitus removed surgically 16 months prior to first test date; no reoccurrence; dry skin antero-distally	antero-distal tissue breakdown medio-distally near osteocyte; notes increased frequency of sores with pistoning	folliculitus at postero-medial trimline; lipoma was removed surgically 9 months prior to first test date; no reoccurrence

dynamic magnitudes are also reported and interpreted.

Methods

Subjects: Test subjects were three active male unilateral trans-tibial amputees between the ages of 23 and 46 who regularly used patellar-tendon-bearing sockets with sleeve suspensions. They were non-diabetic, had no diagnosed neurological problems, and had no history of peripheral vascular disease. In the period 6 months before the clinical studies and during the clinical studies, no major stump skin breakdown problems occurred. For all subjects the amputated leg was the right leg. Descriptions of subject stumps are given in Table 1.

Instrumented prosthesis: All prosthetic design, fabrication, and fitting was performed by certified prosthetists.

A "total contact" hard socket and Pelite® liner were designed for each subject using Seattle ShapeMaker© (Prosthetics Research Study, Seattle, Washington) software and a vacuum-forming socket fabrication system (Davies and Russell, 1979). Sockets and liners were designed to be slightly smaller than those normally worn by a subject since in the interface stress studies no socks or nylon sheaths were worn between the stump and socket. Several design iterations were usually required before a fit was determined appropriate.

Normal and shear interface stresses were measured during walking trials. Custom-designed transducers (Sanders and Daly, 1991) that measured stresses in three orthogonal directions were positioned in mounts bonded to the external socket surface. Transducers protruded through holes in the socket and Pelite liner so that their sensing surfaces were flush with the inside liner surface. Because the transducer surface was made of Pelite, no foreign material was introduced to the stump. Forces and moments in the prosthetic shank were measured simultaneously using instrumentation described elsewhere (Sanders, 1991).

Sites for interface stress monitoring were selected. Clinical locations of interest on the anterior, posterior, and lateral surfaces were used. The tibial crest, patellar tendon, and fibular head regions were avoided because their high curvatures would have caused improper function of the instrumentation. Medial site locations were not used because of interference by the contralateral limb. Brim locations were avoided since the mounts would have interfered with the function of latex sleeve suspension. Example transducer sites for a subject are shown in Figure 1a. The seven regions were the same for all subjects except Subject #3 where the antero-medial distal site was located further proximal because of scar tissue clefts in the skin. In Subject #1 the lateral site was not tested. Only four of the seven sites were monitored in a data collection session because of limitations of the data acquisition system. Sites that were not monitored were filled with dummy transducer plugs to ensure that the pressure difference across the socket wall was maintained. An instrumented socket is shown in Figure 2.

The socket was bonded to a wood block using standard methods and materials (polyester resin and cellulose filler). To complete the prosthesis a Berkeley adjustable leg with an instrumented shank (Sanders, 1991), a Seattle™ Lite Foot (Model and Instrument Development (M+IND), Seattle, Washington) and a latex sleeve suspension were attached. Subject #2 used an elastic waist belt suspension in addition to the latex sleeve. The total mass of each instrumented prosthesis was approximately 3.2kg. The mass of the subjects' normal prostheses, which were thermoplastic limbs with Seattle™ System components, was approximately 1.5kg.

Signal conditioning was performed by equipment housed within a backpack box carried by the subject. A heavily-shielded cable extended from the backpack box to the prosthesis while a second thin cable extended from the backpack box to a computer data storage facility. The cables did not restrict a subject's normal range of motion but they did add mass to the prosthesis. The mass of the backpack box and cable was approximately 3.2kg.

Using standard techniques, static and dynamic

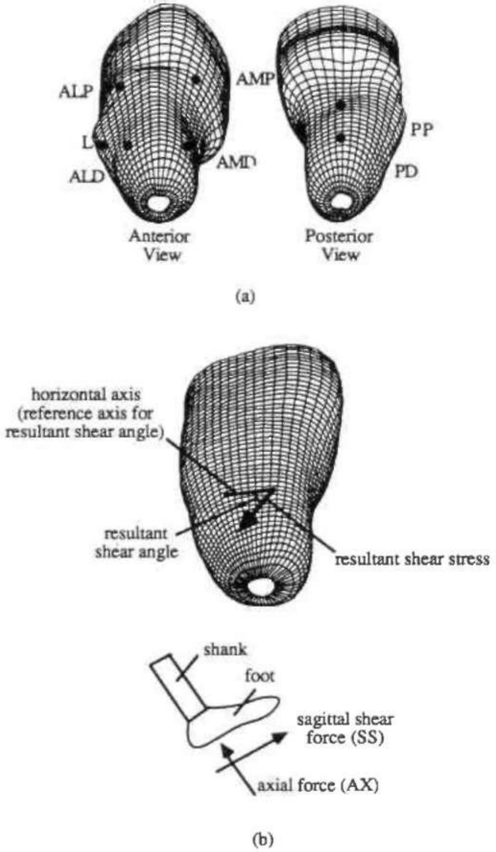

Fig. 1. (a) Transducer sites for Subject #2 are shown. ALP=antero-lateral proximal, AMP=antero-medial proximal, ALD=antero-lateral distal, AMD=antero-medial distal, L=lateral, PP=postero-proximal, PD=postero-distal. (b) "Resultant shear stress" is in the plane of the interface. Resultant shear angles are referenced to a horizontal axis and they increase in a counterclockwise direction. An approximately +45 degree resultant shear angle is shown. In the prosthetic shank, axial force is along the pylon axis and sagittal shear force is in the shank cross-section in a sagittal plane.

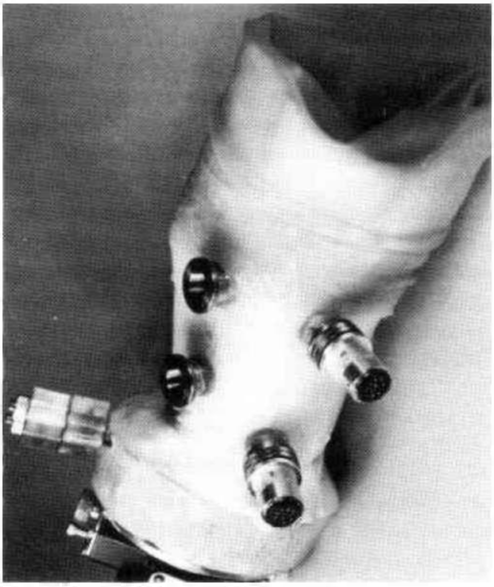

Fig. 2. Transducers occupy antero-medial sites. A strain-relief clamp for the cables (not shown) is affixed to the wood block at the lower-left.

prosthetic alignment were performed by certified prosthetists. The selected socket/shank alignment setting was defined as the "zero" or optimal alignment. "Zero" alignment was the same for all sessions conducted on a subject.

Testing protocols: Both walking and standing trials were conducted.

The pathway for walking was a 1.2m × 18m hallway that had a concrete floor covered with vinyl tiles. Data collection began ½ to 1 step after walking was initiated and lasted for 8 seconds. Usually 6 to 7 steps were collected in a trial though only the central 4 to 5 complete steps were used in analysis.

Spectral analysis of interface stress data collected at 500Hz sampling rate showed signal bandwidth to be less than 40Hz. In subsequent walking trials a 125Hz sampling rate with appropriate anti-aliasing filtering was used.

Trials were conducted with sagittal plane angular alignment at one of three settings: "zero", "plantarflexion", or "dorsiflexion". "Zero" was optimal alignment as determined by a team of prosthetists. "Plantarflexion" was an angular change of approximately 9 degrees with respect to an axis through the Berkeley jig and perpendicular to the sagittal plane. "Dorsiflexion" was an angular change of approximately 6 degrees. A group of at least four consecutive trials was conducted at each setting, thus the alignment was changed twice over the course of a data collection session. Sessions were approximately 30 minutes in length. The order of alignment setting was randomly selected.

Walking rate was controlled to a value between 94 and 99 steps/min using a metronome. The value selected for each subject was approximately his normal walking speed. Mid-trial stride length was measured during three of the trials for each subject using reference markers on the floor. Walking velocities for all subjects were approximately 1.3m/s.

Standing trials were conducted at the beginning and at the end of each session. During standing trials, a subject was asked to achieve one of four weight-bearing levels: (i) full weight-bearing, (ii) equal weight-bearing, (iii) low weight-bearing, or (iv) no weight-bearing. Prosthetic shank force data displayed on the computer monitor was used for feedback to the subject. Data was collected approximately 5 seconds after the subject had achieved a stable position. Results were not dependent on the order of weight-bearing tests.

The number of sessions conducted for Subjects #1, #2, and #3 were two, three, and four respectively.

Results

The following conventions are used in data presentation. Results are for "zero" alignment unless otherwise stated. By definition "normal stress" is perpendicular to the interface. "Resultant shear stress" is orthogonal to normal stress and is the resultant stress component in the plane of the interface (Fig. 1b). Compressive forces on the transducers are positive in sign. Resultant shear angles during the central portion of stance phase are in block parenthesis on the right side of interface stress plots. Stresses and angles are referenced to a transducer coordinate system as shown in Figure 1b.

Walking

All subjects loaded their prosthetic leg during approximately 62% of each step cycle. The average prosthetic stance phase duration for all subjects was 0.79s and the average step duration was 1.26s.

At a site, shapes of interface stress wave-forms changed from one step to another, probably as a result of compensation. However, repeated characteristics were evident in data from different steps. They are described in this section.

Interface stress curves during stance phase were divided into seven sections for analysis as shown in Figure 3. All sections were not necessarily present in all wave-forms and their timings were not necessarily as shown in Figure 3. This figure is intended only as a section-identification reference.

Loading delays: Stresses at transducer sites did not begin to increase immediately at heel contact. For Subjects #1 and #2 there was a delay of between 5% and 12% of stance phase

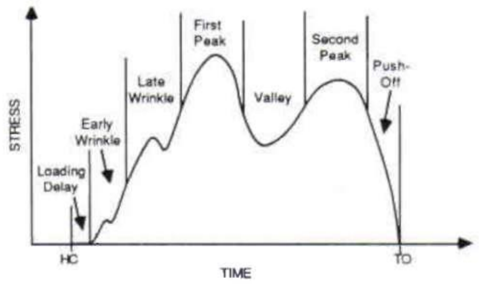

Fig. 3. The stance phase of interface stress wave-forms was divided into seven sections for analysis. HC=heel-contact, TO=toe off.

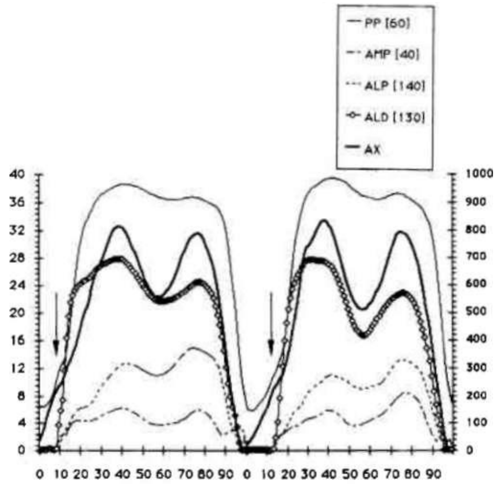

Fig. 4. Resultant shear stresses and shank axial force (AX) for stance phase of two consecutive steps for Subject #1 are shown. The swing phase has been removed from the record. Arrows indicate the ends of "loading delays" for antero-proximal sites. Numbers in block parentheses in the figure legend are resultant shear angles during the central phase of stance. The left scale is normal stress in kilopascals (kPa) and the right scale is shank shear force in Newtons (N). 1kPA=0.14lb/in^2, 1N=0.22lb. PP=postero-proximal, AMP=antero-medial proximal, ALP=antero-lateral proximal, ALD=antero-lateral distal.

(approximately 0.04s to 0.10s) while for Subject #3 the delay was approximately 4% (0.03s). As shown in Figure 4, for Subject #1, axial force in the prosthetic shank could reach as high as 25% of its maximal value when anterior site interface stresses began to increase. Usually "loading delays" were longer at anterior sites compared to posterior sites.

High freqency events: In most of the steps there were "high frequency events" (HFE's) in interface stress curves soon after heel contact. "Early HFE's" were high frequency spikes at about 7% to 12% into stance phase. "Late HFE's" were of lower frequency content and occurred later, usually partway up the rising part of interface stress curves.

Early HFE's: "Early HFE's" were apparent in both normal and shear stress wave-forms but they were not always present at all sites. If they occurred in a given step they happened at approximately the same time at different sites and their timings matched up well with HFE's in the shank wave-forms, particularly axial and sagittal shear forces (Fig. 5).

Late HFE's: "Late HFE's" were not as clearly defined as "early HFE's" in that they did not all occur simultaneously in a given step (Fig. 6). Nor were the times at which they occurred at a site consistent from one step to another. They could occur early and separate from the rest of the curve or they could occur late near the first peak and be virtually buried under that peak (Fig. 6b). However, they consistently occurred before the shank resultant force changed from being directed anteriorly to being directed posteriorly (Point "A" in Fig. 10a).

The ratio of resultant shear stress to normal stress was investigated to determine if there was a threshold value for the occurrence of "late HFE's". No consistent value was found at any site. Resultant shear angles showed no maxima, minima, or inflection points during "late HFE's".

First peaks: Peaks in interface stress wave-forms usually occurred near the maxima of the absolute values of axial force, shear force, and sagittal bending moment about the pylon centre in the prosthetic shank but they were not necessarily within a window surrounding those peaks (Fig. 7). Also, if distorted by the presence of "late HFE's", they could occur earlier. An example is the second step in Figure 6a at the postero-proximal site.

Timings of maxima at different sites in the same step were not simultaneous nor were the timings of maxima at the same site in different directions simultaneous. For example in Figure 7 antero-medial distal normal stress reaches a maximum

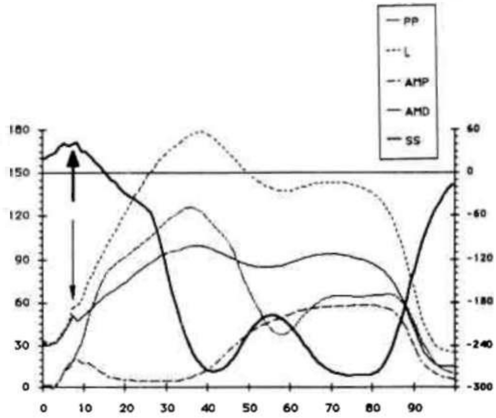

Fig. 5. Timings of "early HFE's" in normal stresses are indicated with a thin arrow for Subject #3. Timing of an "early wrinkle" in shank sagittal shear force (SS) is indicated with a thick arrow. The left scale is normal stress in kilopascals (kPa) and the right scale is shank shear force in Newtons (N). 1kPa=0.14lb/in^2, 1N=0.22lb.

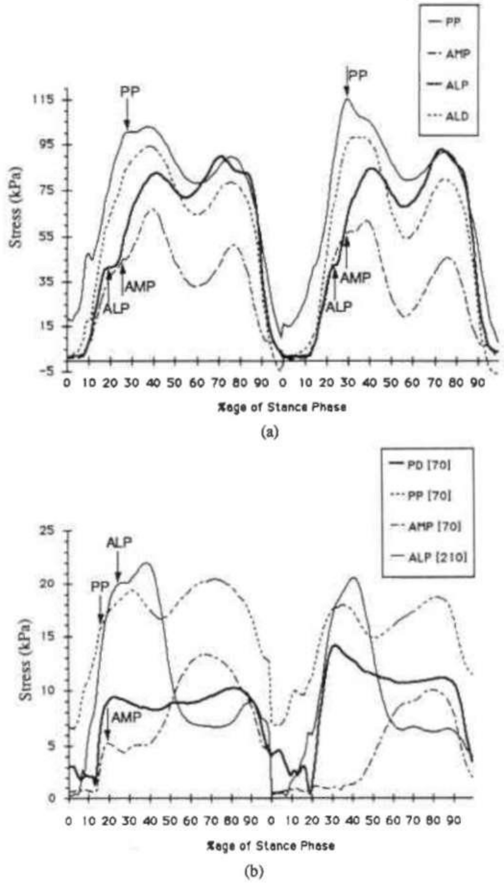

Fig. 6. (a) Normal stresses for stance phase of two consecutive steps for Subject # 1 are shown. Late HFE's are indicated by arrows. PP=postero-proximal, AMP=antero-medial proximal, ALP=antero-lateral proximal, ALD=antero-lateral distal. (b) Resultant shear stresses for stance phase of two consecutive steps for Subject #2. PD=postero-distal, PP=postero-proximal, AMP=antero-medial proximal, ALP=antero-lateral proximal.

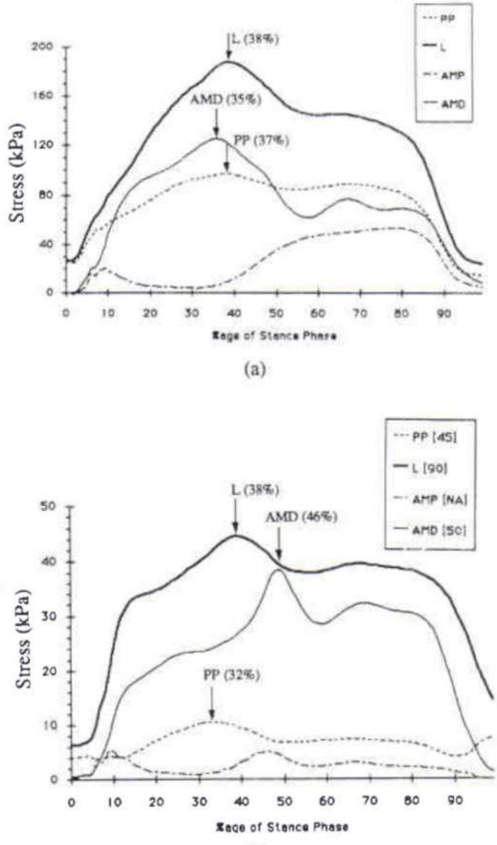

Fig. 7. (a) Normal stresses and (b) resultant shear stresses during stance phase for Subject #3. Data are from the same step. Shank axial force and sagittal shear force first peaks occur at 38% and 41% respectively. PP=postero-proximal, L=lateral, AMP=antero-medial proximal, AMD=antero-medial distal.

slightly earlier than either postero-proximal normal stress or lateral normal stress. Antero-medial distal normal stress peaks earlier than antero-medial distal resultant shear stress.

Valley and second peaks: Valley and second peak regions were of lower frequency content compared with earlier in stance. An absolute value shear force/axial force ratio of approximately 0.2 to 0.4 corresponding to a resultant shear angle range of $-11°$ to $-21°$ was maintained throughout the valley and second peak regions (Fig. 10). The axial force on the bottom of the foot stayed at approximately the same sagittal plane position (Fig. 8).

Push-off: Interface loading immediately before toe-off was not simply a reversal of that immediately after heel contact. For all subjects there were usually no "HFE's" or "loading delays" on the unloading phases of interface stress wave-forms. Ordering of loading did not necessarily match ordering of unloading. Unloading resultant shear vectors were not opposite in magnitude and direction from loading resultant shear vectors.

Different alignments

Changes in angular alignment did not significantly affect stance durations, step durations, or stance/step ratios. Thus the subjects still maintained their cadence well with the metronome despite the modifications.

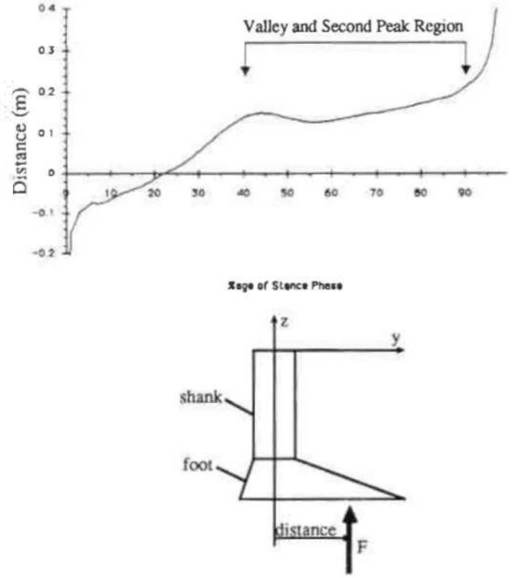

Fig. 8. Distance of the force "F" from the pylon axis is shown for Subject #3. "F" is parallel with the pylon axis.

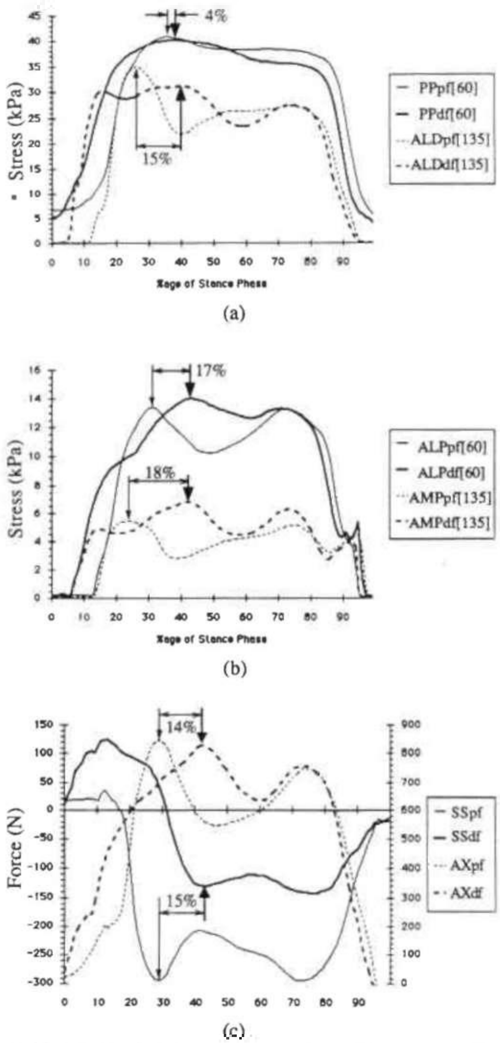

Fig. 9. (a) Resultant shear stresses for postero-proximal (PP) and antero-lateral distal (ALD) sites for "plantarflexion" (pf) and "dorsiflexion" (df) alignments. (b) Resultant shear stresses for antero-lateral proximal (ALP) and antero-medial proximal (AMP) sites. (c) Shank axial force (AX) and sagittal shear force (SS). All data are from the same step of Subject #1.

Interface stress "loading delays" and "early HFE's" occurred at approximately the same time as for "zero" alignment. "Late HFE's" still occurred before the shank resultant angle in the sagittal plane became negative.

The times at which peak stresses occurred did change. For Subject #1 and to a lesser degree for Subject #3, peaks in the first half of stance occurred earlier for the "plantarflexion" setting than for the "dorsiflexion" setting (Fig. 9). For Subject #2 the opposite trend was found. Peaks occurred later for "plantarflexion" than for "dorsiflexion". Resultant force directions in the sagittal plane remained approximately the same for different alignment settings (Fig. 10).

Mid-stance interface stress wave-forms, in general, shifted to the left for "plantarflexion" changes and shifted to the right for "dorsiflexion" changes. These shifts were usually not more than a few percent of stance phase.

Standing vs. walking

In walking trials, in general, antero-lateral distal and antero-medial distal sites were loaded during stance and unloaded during swing while postero-proximal, postero-distal, lateral, antero-lateral proximal, and antero-medial proximal sites were loaded during both stance and swing (Fig. 11). Standing trial data showed related patterns. Sites loaded during standing with equal or full weight-bearing were usually loaded during the stance phase of gait. Sites loaded during standing with minimal weight-bearing matched well with swing phase loaded sites. As shown in Figure 11 there were some exceptions to these trends. For example, Subject #3 unloaded the antero-lateral proximal site during stance but loaded it during swing. Subject #1 unloaded all anterior sites during swing.

Under equal weight-bearing on both feet or full

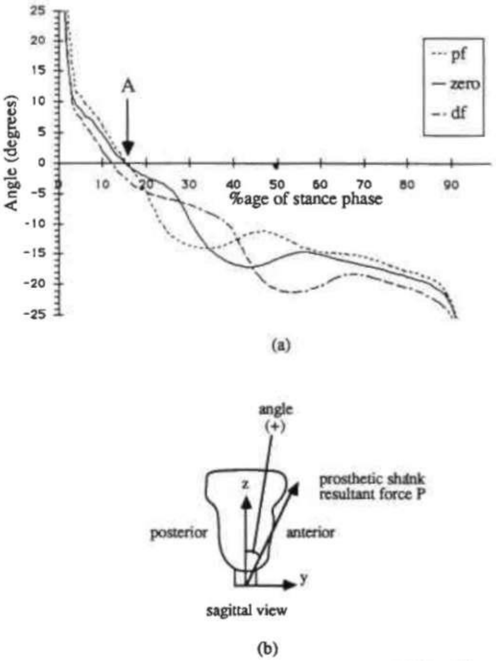

Fig. 10. Resultant force measured with the instrumented shank is shown for stance phase from steps at three alignment settings for Subject #3, pf="plantarflexion", df="dorsiflexion". At point "A" the resultant force for "zero" alignment changes from being directed anteriorly to being directed posteriorly.

weight-bearing on one leg, usually antero-distal sites were loaded more than antero-proximal sites. Ratios of resultant shear stress to normal stress usually stayed approximately the same or decreased from equal weight-bearing on both feet to full weight-bearing on one leg at all sites except at some antero-proximal locations.

Stance phase peak normal stresses averaged 2.4($\pm$0.6) times stresses achieved during standing with equal weight-bearing on both feet and 3.3($\pm$2.5) times stresses achieved during one-legged standing with full weight-bearing. Stance phase resultant stresses averaged 1.4($\pm$0.6) times stresses achieved during standing with equal weight-bearing on both feet and 2.1($\pm$2.0) times stresses achieved during one-legged standing with full weight-bearing.

Discussion

To the authors' knowledge this is the first research investigation of interface shear stress measurements on trans-tibial amputee subjects. It is also the first time both normal and shear stresses have been measured simultaneously.

In interpreting these data three characteristics of the instrumented prostheses which made them different from subjects' normal prostheses should be recognised. (i) Instrumented limbs were

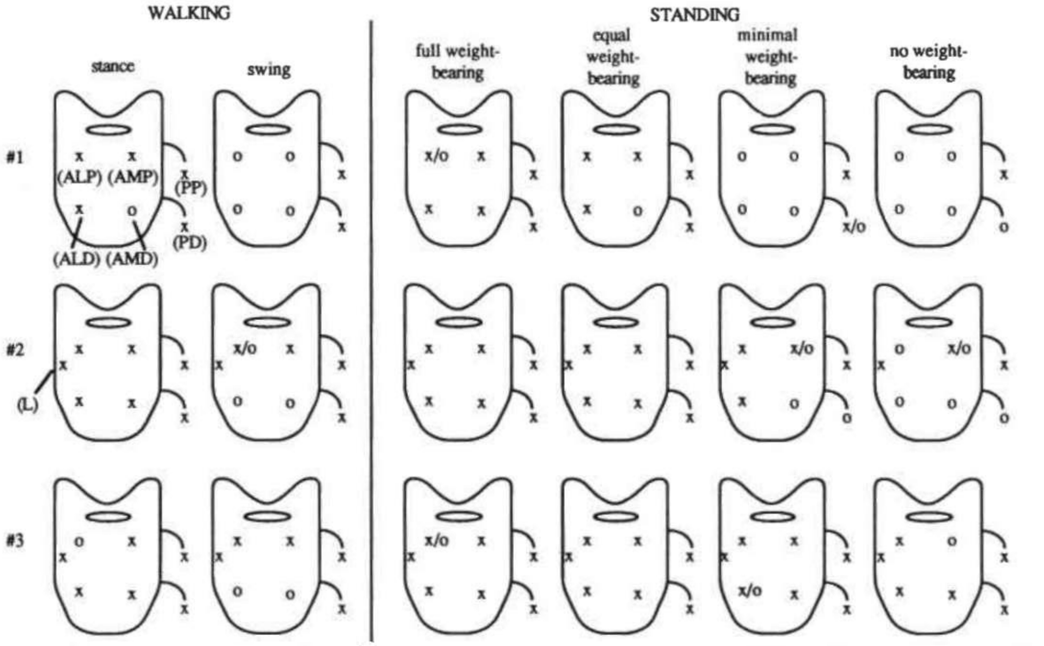

Fig. 11. Results for Subjects #1, #2, and #3 are shown. A site was considered loaded during stance or swing phase if it experienced greater than 5kPa stress during at least 50% of the phase. x=loaded, o=unloaded, x/o=loaded in some but not all of the sessions.

heavier than subjects' normal prostheses. (ii) No sock was worn. (iii) A 3.2kg backpack and cable apparatus was worn.

Loading delays: "Loading delays" were probably a result of one or both of the following sources: (i) "pistoning", i.e. slip between the stump and the socket. (ii) the anteriorly-directed resultant shank force (force "P" in Fig. 10b).

Clinical relevance of "loading delays" is that at the end of the delays the position of the stump in the socket is probably set for the rest of stance. If this position is different from that set under clinical fitting conditions, there may possibly be undesirable contour mismatches between the stump and prosthetic socket surfaces. Stress concentrations would be induced which could traumatise stump tissues.

Time lengths of "loading delays" may possibly also be important. If delay phases in longitudinal shear stresses end at unequal times at different sites, then intermediate skin tissue would be exposed to in-plane tension in a longitudinal direction. High in-plane tension can cause skin blanching and obstruction of blood flow (Kenedi *et al.*, 1965). Tension can also change the mechanical response of skin in the perpendicular direction, i.e. the transverse shear direction. Biaxial mechanical testing of rabbit skin has shown that high perpendicular stretch ratios cause the stress-strain curve to shift towards the origin (Lanir and Fung, 1974). This means that the stress level for blanching would happen at a lower uniaxial strain and at a lower internal energy change than if perpendicular strains were zero. Presumably the potential for tissue trauma would also be increased. As shown in this research, in stance phase immediately subsequent to "loading delays", transverse shear was usually greater than longitudinal shear on the anterior limb surface, thus it is possible that biaxial loading was induced.

Early HFE's: "Early HFE's" could have been due to several sources: (i) the foot being forced to foot-flat, (ii) deformation of the foot, (iii) activity of the other leg, (iv) muscle activation, or (v) slip at the interface. Because interface stress "early HFE's" matched well with "early HFE's" in the prosthetic shank, (i) or a combination of (i), (ii), (iii), and (iv) is probable.

If due to the foot being forced to foot-flat, deformation of the foot, or activity in the other leg, then "early HFE's" may possibly affect an amputee's sense of stability but they probably have minimal effect on tissue mechanics. Unless the skin is highly sensitive, tissue damage due to high-frequency low-intensity loading is unlikely (Naylor, 1955). In mechanical testing, skin response has been shown to be virtually insensitive to strain rate in the range of interest here (Lanir and Fung, 1974).

However, if "early HFE's" were due to muscle activation, then they may be providing important information early in stance. Muscle activation would be expected to change stump muscle geometry and stiffness. Thus "early HFE's" could serve as a signal for this change which could be important in subsequent interface stress generation. Knowledge of changes in muscle geometry and material properties could also be important when attempting to model mechanically the stump (Steege *et al.*, 1987; Krouskop *et al.*, 1987; Quesada and Skinner, 1991; Sanders, 1991; Torres-Moreno *et al.*, 1992) for the purpose of predicting interface stresses for use in socket design.

Late HFE's: It is possible that the principal source of "late HFE's" was the change in direction of shank sagittal shear force from being directed anteriorly to being directed posteriorly (Fig. 10). The bending moment on the tibia reversed direction and, because the bone was imbedded in soft nonlinear tissue, "HFE's" appeared in interface stress curves.

Clinical relevance of "late HFE's" is that, at these load levels, a quick interface stress reduction followed by a quick stress increase may possibly induce tissue damage. Friction blister studies (Naylor, 1955) have shown that blister occurrence is related to the frequency of the applied load cycles. Thus interface stress waveforms with "late HFE's" may possibly be more detrimental than those without "late HFE's". Also, because "late HFE's" did not all occur simultaneously in a step, it is possible that stress concentrations were induced in soft tissues.

First peaks: First peaks in interface stress waveforms did not usually occur simultaneously with each other nor did they necessarily occur simultaneously with shank force and moment maxima. This may, in part, be due to the nonlinear viscoelastic nature of stump tissues.

The lack of simultaneously-occurring peak interface stresses is significant because the tissue mechanics of the stump is affected. Peaks not occurring simultaneously in resultant shear stresses at different sites means that tensile stresses may possibly occur between loaded sites. At regions where skin is adherent to underlying bone, concentrated shear stresses would occur adjacent to or within adherent scar tissue. This could be particularly detrimental on the anterior surface where skin over bone is thin.

Valleys and second peaks: Mid- and late-stance interface stress curves lacked the characteristic features of early stance phase.

Push-off: There may possibly be important tissue mechanics manifestations of differences between heel-contact and push-off. Because wave-forms were not symmetrical, skin experienced resultant shear stress in principally one direction. Fluids were pushed unidirectionally, which changes their distribution beneath the skin surface. For example, at the antero-distal end, fluid was possibly pushed proximally and away from the midline of the tibia but not in the reverse directions. This may possibly explain why dry skin is sometimes seen at antero-distal ends of stumps. Subject #1 had particularly dry skin in the antero-distal area (Table 1).

Different alignments

Changes in interface stress wave-form shapes for "plantarflexion" and "dorsiflexion" alignment modifications reflect compensation. Subjects changed their gait to achieve stability and comfort. Wave-form shape changes were not consistent across transducer sites or across subjects, however, thus each subject compensated his gait style differently. However there were two parameters that consistently did not change for all subjects. They were: (i) ranges of shank resultant force directions in the sagittal plane (Fig. 10) and (ii) interface stress peak magnitudes (Sanders, 1991).

All subjects in this research were well-conditioned amputees. Possibly to maintain their usual sense of stability and comfort, amputee subjects kept ranges of shank resultant force directions and interface stress peak magnitudes consistent. New amputees however might not have that conditioning. Thus consistent angles and interface stress peaks would possibly not be achieved when alignment modifications are made during first-time fittings.

Results presented here are consistent with work by Pearson *et al.* (1973). Interface pressures at the patellar tendon, the lateral tibial condyle, and the medial tibial condyle stayed approximately the same for 10 degrees of flexion or extension alignment modification. Translational antero-posterior alignment changes were shown to cause changes in interface pressure wave-form shapes.

Standing vs. walking

Results showing loss of contact for swing phase and for low or minimal weight-bearing conditions are expected. It is likely that "pistoning" occurred at the interface. Exceptions such as the antero-lateral proximal site on Subject #3 and the antero-medial distal site on Subject #1 show that despite the intended socket design, sockets were not "total contact".

Ratios of "equal weight-bearing (symmetrical standing) stress: maximal stance phase stress" showed lower standard deviations than ratios of "full weight-bearing (one-leg standing) stress: maximal stance phase stress". Thus equal weight-bearing was a more accurate reflection of dynamic stress distribution during peak loading than was full weight-bearing. However, it should be noted that stress distribution differences between equal weight-bearing and dynamic loading were still rather substantial thus it would not be appropriate to conclude that there was a single ratio relating them. Standard deviations/averages for ratios of "equal weight-bearing stress: maximal stance phase stress" were 25% and 43% for normal and resultant shear stress respectively. It is likely that the high sagittal plane bending moment and the more posteriorly-directed shank resultant force vector achieved during gait compared to those achieved during standing were responsible for the differences.

Further research

Further research will be in two directions. First, interface stress measurement instrumentation will be modified so that the weight of the instrumented prosthesis is reduced, the backpack box is eliminated, a stump sock or sheath can be worn, and more transducers can be monitored simultaneously. The instrumentation will be used to investigate other parameters including: alignment changes in other directions, effects of socket shape modifications, and different prosthetic feet. A principal goal is to understand interface stress sensitivity to changes in prosthetic design. Another goal is to understand amputee adaptation to prosthetic modifications. In the long-term, knowledge of sensitivity and adaptation will be integrated into computer-aided prosthetic design.

A second direction of research is to investigate skin response to normal and shear loads. The intent is to better understand how skin adapts to mechanical stress to become durable and load-tolerant. Knowledge gained will be applied to design of rehabilitation treatment strategies that encourage tissue adaptation and minimise risk of tissue breakdown.

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