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Mechanical properties of prosthetic limbs: Adapting to the patient

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Abstract—Lower-limb amputees have identified comfort and mobility as the two most important characteristics of a prosthesis. While these in turn depend on a multitude of factors, they are strongly influenced by the biomechanical performance of the prosthesis and the loading it imparts to the residual limb. Recent years have seen improvements in several prosthetic components that are designed to improve patient comfort and mobility. In this paper, we discuss two of these: VSAP and prosthetic foot-ankle systems; specifically, their mechanical properties and impact on amputee gait are presented.

Key words: *amputation, artificial limbs, biomechanics, pain, rehabilitation, residual limbs.*

INTRODUCTION

The primary goal of rehabilitation of the individual who has undergone lower-limb amputation is the optimum restoration of function. Although many aspects of the rehabilitation plan—beginning with the surgical care, physical therapy, occupational therapy, and psychological support—can influence the rehabilitation outcome, one of the most important determinants is the quality of the prosthetic restoration. This depends primarily upon the fit of the prosthetic socket and the choice of prosthetic components. The physician and prosthetist who formulate a prosthetic prescription must integrate marketing and scientific data to make important, cost-effective decisions. Unfortunately, the marketing strategy of most prosthetic manufacturers is to persuade providers and patients to choose the most expensive components, and the scientific evaluation of these components is frequently quite limited. As we formulate a prosthetic prescription, we must be simultaneously aware of the prosthetic characteristics that are most important to the amputee and of the characteristics that will most strongly influence their functional status.

Patient comfort is one of the most important factors that influence the functional status a patient ultimately achieves. In a survey of veterans and nonveterans with lower-limb amputations, Legro et al. (1) found that the fit and comfort of the prosthesis, and the avoidance of blisters or sores on the residual limb, were the most important functional characteristics of the prosthesis. Similarly,

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Postema et al. (2) found that amputees rate "absence of stump pain" and "no fatigue during walking" as the most important subjective aspects of a prosthesis. Although its negative impact on function is widely appreciated, residual-limb pain continues to be a pervasive problem for the amputee. In a survey of 255 amputees, Ehde et al. (3) found that 74 percent had residual-limb pain with a mean intensity of 5.4 on a 0 to 10 scale; further, of those with pain, 60 percent described it as moderately to severely bothersome.

The factors responsible for an amputee's perception of discomfort have not been firmly established but are likely to include abnormal loading of the residual-limb soft tissues, abnormal or excessive musculoskeletal loading of the residual-limb proximal structures, and/or abnormal or excessive loading of the intact limb musculoskeletal structures. In the prosthetic lower limb, the forces associated with the impact at heelstrike, the support of body weight, the acceleration and deceleration of the center of mass, and the swing phase must be transmitted through the soft tissues of the residual limb and ultimately through its skeletal and musculotendinous structures. The soft tissues of the residual limb are not adapted to load bearing and in many situations may be compromised by scar tissue, split thickness skin grafts, or neuromata.

A number of prosthetic-limb characteristics may influence tissue loading. These include the quality of the suspension system which, if poor, can result in residual-limb "pistoning"; the quality of the socket design and contours; the type of interface material that is interposed between the hard socket and the tissues of the residual limb; the ability of the prosthetic pylon to absorb or dissipate forces; and the mechanical properties of the prosthetic foot, the most important of which are the stiffness of the heel and forefoot keel. Choosing appropriate components as part of the prosthetic prescription, while optimizing socket contours and prosthetic alignment, is critical to optimum user comfort.

Although secondary to comfort, the degree of mobility provided by the prosthesis is extremely important to functional outcome. In the survey by Legro and colleagues (1), the importance of maintaining mobility was rated 97.6 on a scale of 0 to 100. Similarly, Kegel (4) found that functional limitations such as the inability to run and excessive fatigue were the main reasons given by amputees for their lack of participation in recreational sports. The level of mobility is largely a function of the biomechanical characteristics of the prosthesis.

To properly evaluate the biomechanical characteristics of a prosthesis, the most important functional requirements of the intact ankle, foot, and associated musculature during normal gait must first be understood. At the instant of heelstrike, the rapid deceleration of the stance limb results in a high-frequency "shock" wave that is transmitted through the skeleton. This impact force is attenuated by passive shock absorbers such as the heel pad, articular cartilage, and synovial fluid, as well as by active absorbers such as joint motion and muscle activity. Following heelstrike, the anterior musculature of the stance leg contracts eccentrically, as do the quadriceps, to absorb the impact forces and to control the vertical deceleration of the body's center of mass. During midstance, eccentric contraction of gastrocnemius-soleus controls the progression of the center of mass. Finally, in late stance phase gastrocnemius-soleus provides the greatest power output of all lower extremity muscle groups; this positive power output is a product of both concentric muscle work done by the muscle complex and the passive return of elastic energy that was stored in it during midstance lengthening. The positive power output may to a small extent accelerate the body's center of mass, but more importantly, it serves to accelerate the leg forward into swing phase. A key feature of the intact musculoskeletal and neuromuscular foot/ankle complex is its adaptability. It can function optimally over a wide spectrum of walking and running speeds as well as over varying terrains.

There have been numerous prosthetic components designed to meet the requirements described above. Most of these have been foot/ankle systems, but more recent years have seen the introduction of "vertical shock-absorbing pylons" (VSAPs). As their name implies, these pylons are designed to attenuate the shock loads generated during walking in an attempt to increase comfort and decrease fatigue. Improving comfort and mobility are also the goal of foot/ankle systems that incorporate energy-storing components. These devices attempt to reproduce the normal elastic energy-storing mechanisms of the gastrocnemius-soleus. Although advances such as VSAPs and energy-storing foot/ankle systems provide significant restoration of function, the positive power output and adaptability of the normal limb have yet to be achieved.

This paper will address two important elements of the transtibial prosthetic prescription that influence comfort and biomechanical function. Specifically, the contribution of vertical shock pylons to tissue and musculoskeletal loading, as well as the effects of the

mechanical characteristics of prosthetic feet on tissue loading and the biomechanics of amputee gait will be discussed.

VERTICAL SHOCK-ABSORBING PYLONS

Tissue Damaging Transient Loads During Walking

During walking, the human body is subjected to repetitive, high-magnitude forces as it comes into contact with the ground. While these ground reaction forces are generally of low to moderate frequency, at the instant of heelstrike, the rapid deceleration of the stance limb results in a high-frequency "shock" wave that is transmitted through the skeleton. It is observed as a short spike of force superimposed on the upslope of the ground reaction force (**Figure 1**). Spectral analysis has shown 99 percent of the power is contained below 15 Hz, although additional transients provide components above 50 Hz (5).

Research suggests that repetitive loading in general, and *high-frequency* repetitive loading in particular, can be harmful to the musculoskeletal system. Such loading has been implicated in the initiation and progression of osteoarthritis, prosthetic joint loosening, low-back disorders, and other "over-use" syndromes (6–10). It has also been associated with soft tissue damage; studies have shown that repetitive loading can lead to inflammatory autolysis of the skin and ultimately to ulceration (11,12).

Because lower extremity amputees clearly lack the majority of "natural" shock absorbers, they are particularly vulnerable to the adverse effects of the repetitive loads experienced during gait. Soft tissue sites unfamiliar to repetitive loading, such as those of the residual limb, cannot easily adapt and are prone to tissue breakdown and localized pain (13). This is reflected in the high frequency and severity of residual-limb pain in this population; as many as 74 percent of amputees suffer from residual-limb pain, and 60 percent consider it moderately to severely bothersome (3).

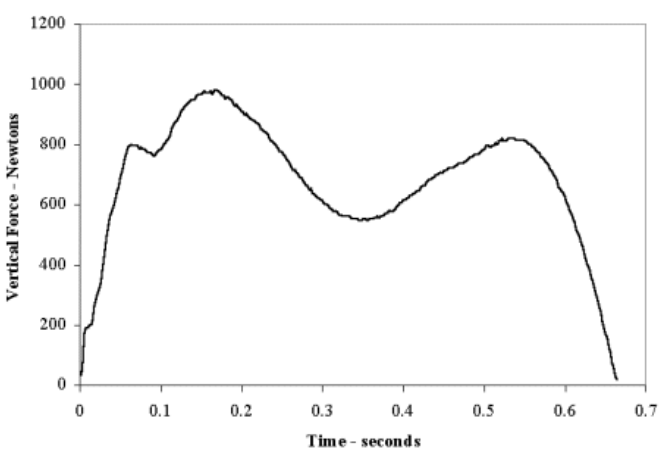

Figure 1
Vertical component of the ground reaction force of a 89-kg male walking with a Flex-Foot prosthesis.

In an attempt to ameliorate residual-limb problems, prosthesis manufacturers have recently marketed VSAPs, pylons designed to attenuate the deleterious loads generated during walking (**Figure 2**). In contrast to traditional rigid pylons, these pylons have a compression element that displaces under weight bearing. The equivalent stiffness and damping of the pylon are determined primarily by the choice of compression element. Despite the commercial availability of these devices, few investigators have attempted to characterize their mechanical properties or evaluate their impact on amputee gait.

Figure 2
Three commercially available VSAP. From left to right: (a) TT Pyramid Pylon (Blatchford Endolite, Chas A. Blatchford & Sons Ltd., Hampshire, U.K.), (b) ICON™ Shock Pylon (Flex-Foot Inc., Aliso Viejo, CA), (c) Total Shock Pylon (Century XXII Innovations Inc., Jackson, MI).

Mechanical Properties of VSAP

Before their effect on amputee gait can be fully understood, the mechanical response of VSAP to standardized loading conditions must be evaluated. To date,

few attempts to do so have been made. Miller and Childress (14) were the first to conduct mechanical testing of VSAP. Data from static and dynamic tests of the Flex-Foot Re-Flex Vertical Shock Pylon were used to calculate the linear time-invariant constants of a second-order mass-spring-damper model. When the assumptions are valid (i.e., linear and time-invariant), this type of model can be used to predict attenuation across the frequency spectrum, including suspected tissue-damaging frequencies. Their calculated spring constant for the pylon was of comparable magnitude to values reported in the literature for the intact limb. Despite this agreement, the authors correctly noted that a linear second-order model is likely an oversimplification, suggesting that it may be of limited use for predicting response to high-frequency vibrations.

Gard (15) expanded the work of Miller and Childress (14) to include vertical shock pylons from three different manufacturers using the same experimental methods. The force *versus* deformation test demonstrated that only one of the three pylons had a linear spring constant, invalidating the use of a second-order linear time-invariant model to compare between pylons. The preliminary work by Gard is particularly valuable because it demonstrates that linear, second-order models are likely to be inaccurate at predicting attenuation at high frequencies. Future work will require methods that address the nonlinear properties of the shock-absorbing pylons.

Effect of VSAP on Amputee Gait

Of ultimate interest to the rehabilitation professional is whether VSAP improve the comfort and function of their amputee patients. Unfortunately, few data exist to guide them in this respect.

To date, only two studies have examined the effect of VSAP on amputee gait (14,16). Miller and Childress (14) tested the Re-Flex VSP on two subjects, with and without the telescoping pylon immobilized. Vertical ground reaction forces, contact times, pylon displacement, and vertical trunk motion were measured for each experimental condition as the subjects walked at a comfortable and at a fast pace, jogged in place, and stepped off a curb onto each limb. Few differences were observed in these variables with the pylon immobilized compared with the pylon functional, suggesting that the addition of vertical compliance does not make an objective difference in amputee gait. However, both subjects expressed a strong subjective preference for the prosthesis with the pylon functional.

Hsu et al. (16) compared differences in metabolic cost, efficiency, and intensity for walking and running with three different prostheses, one of which included a VSAP (i.e., the Re-Flex VSP). They found that for all measures and both modes of locomotion, the prosthesis with a VSAP significantly outperformed the other two. No significant differences were found between the two prostheses without VSAP, despite the fact that one included an energy-storing foot.

Together, the studies by Miller and Childress (14) and Hsu et al. (16) suggest that while VSAP may not significantly affect the mechanics of amputee gait, they may significantly affect the energetics and perception of comfort during amputee gait. This conclusion is tentative, however, and requires further study. As yet, no conclusions can be drawn regarding VSAP's ability to attenuate impulsive forces, since no study has focused specifically on the heelstrike event.

PROSTHETIC FEET (FOOT/ANKLE)

Another aspect of the prosthetic prescription that strongly influences comfort and function is the stiffness of the foot-ankle system. Selection of an appropriate stiffness is primarily based on a patient's body weight, his/her choice of activities, and the intensity level of those activities, but other factors such as residual-limb length, residual-limb pain, and patient sense of stability may also be considered. Nearly all manufacturers allow the physician (or prosthetist) to choose a desired stiffness from a variety of available components. For example, the Seattle Foot (Seattle Orthopedics Group, Poulsbo, WA) is available in seven different keel stiffnesses, and the Flex-Foot (Flex-Foot Inc., Aliso Viejo, CA) has nine categories of stiffness. The physician must weigh the various factors and choose a single stiffness to serve all conditions a patient might experience (see **Figure 3**).

An inappropriate choice of stiffness may lead to increased metabolic costs, abnormal muscle activation patterns, decreased gait symmetry, tissue damage associated with abnormal residual-limb and intact-limb loading, and pain (see, e.g., 2,17–23). Each of these studies describes the biological response to prosthetic limbs with different stiffness profiles and establishes an empirical link between a particular commercial product with an inherent but often unknown stiffness and the observed effects on metabolic cost, muscle activation patterns, gait symmetry, and limb loading.

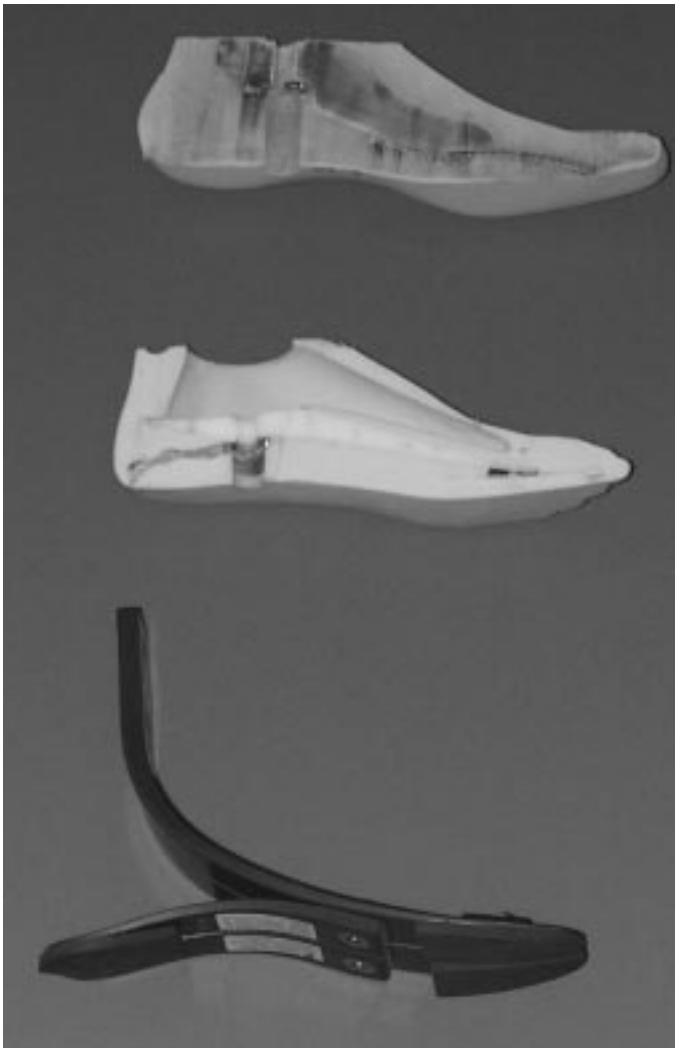

Figure 3
Three commercially available prosthetic feet that exhibit markedly different designs and stiffnesses. From top to bottom: (1) SACH foot (Kingsley Manufacturing Co., Costa Mesa, CA), (2) Seattle foot (Seattle Orthopedics Group, Poulsbo, WA), and (3) Vari-Flex™ foot (Flex-Foot Inc., Aliso Viejo, CA).

In the last 10 years, a number of researchers have conducted investigations to explore the properties of transtibial limbs and how the products from various manufacturers influence amputee gait. Investigators have reported on both the mechanical properties of transtibial limbs in isolated tests and how these properties affect biomechanical and energetic outcomes of amputee gait.

Mechanical Properties of Prosthetic Feet

The stiffness characteristics of the prosthetic limb are designed to substitute for the loss of the muscles and tendons of the intact, biological limb. Whereas the biological limb can vary joint stiffness by altering the extent of muscle activation, the prosthetic limb has no such capability.

In understanding the limitations of these fixed-stiffness prosthetic limbs, most investigators have simply identified the manufacturer of the limb(s) they are investigating. However, a few have measured the properties of the limb that directly influence performance (20,21,24). Lehmann and coworkers (20,21) observed that limb stiffness had a direct influence on lower-limb kinematics, ground reaction forces, joint moments, and step length. To document the differences between the SACH, Seattle, and Flex-Foot feet, they measured the force *versus* deflection characteristics using a quasi-static loading apparatus. From these data, the torque versus angular deflection curve can be calculated (**Figure 4**). The results indicate that there are clear differences in stiffness between feet as well as within feet. All three feet demonstrated monotonically increasing torque as a function of joint angle. During dorsiflexion (positive torque by convention), the SACH foot was the stiffest, followed by the Seattle foot, and then the Flex-Foot foot. During plantarflexion (negative torque by convention), the Seattle foot was the stiffest, followed by the SACH foot, and then the Flex-Foot foot.

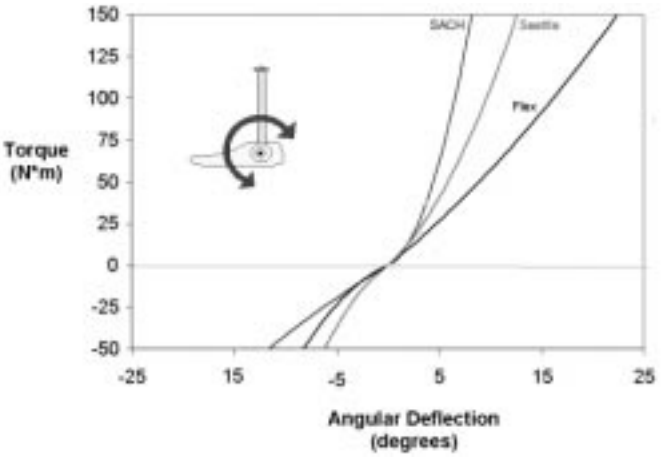

Figure 4
Torque versus angular deflection curves revealing differences in stiffness between the SACH Foot, Seattle Foot, and Flex-Foot (data adapted from reference 21).

Effect of Prosthetic Feet on Amputee Gait—Biomechanical Effects

It is well accepted that the mechanical properties of a prosthetic limb will influence the wearer's ability to

walk or run. However, in a continuous process such as gait, there are an infinite number of variables that might yield an observable response due to a change in a limb property such as stiffness. From the many possible variables, Cortes and coworkers (25) identified 18 plausible kinetic and kinematic variables during the gait cycle that might be affected by the type of prosthetic foot. They performed a statistical analysis of over 1,300 trials of eight traumatic transtibial amputees walking on four different feet and found that prosthetic foot type had a significant impact on several variables that describe lower-limb movement. Most significant was ankle position immediately following heelstrike; their metric was the peak dorsiflexion angle. Large dorsiflexion angles can prolong the period of heel-only support and delay achievement of the stable, foot-flat position. Perry and coworkers (26) studied 10 transtibial amputees wearing three different feet and observed that heel stiffness had a strong influence on the duration of heel-only support prior to foot flat. They found that all three feet contributed to a delayed foot-flat position and concluded that to improve walking performance, the properties of prosthetic feet must promote early foot flat and preserve stability. Prosthetic foot stiffness is clearly a factor in achieving the stable, foot-flat position.

The second most significant kinematic variable of Cortes's study (25) was knee position at mid-stance. In a seminal paper, Saunders et al. (27) identified knee flexion and extension during stance as the third of six gait determinants; knee range of motion is an essential element of normal locomotion. The range of knee flexion during stance in amputees is generally half the range of motion of the intact population (7 *versus* 15 degrees; 28). Tibial progression, and hence knee flexion, can be inhibited by excessive stiffness in a prosthetic foot. In addition to prolonged heel-only support, Perry et al. (26) also found that all three prosthetic feet contributed to reduced knee flexion during level walking. Likewise, Torburn and coworkers (29) observed an inability to advance the tibia and body weight over the foot when their transtibial amputee subjects wore a stiff foot during stair climbing, but not while wearing a more flexible foot. These studies all confirm that foot stiffness influences biomechanics and further work is necessary to better quantify foot stiffness.

Investigators have also looked at the effects of the prosthetic foot type on the intact limb based on the hypothesis that altered gait patterns resulting from amputation are responsible for early degenerative changes observed in the intact limb of unilateral amputees. In a study of 10 traumatic transtibial amputees wearing five different prosthetic feet, Powers and coworkers (30) found prosthetic foot type did influence the impact forces of the sound limb. The Flex-Foot foot significantly reduced the initial peak of the vertical ground reaction force, while the SACH foot consistently produced the greatest ground reaction force. Snyder and others (31), conducting tests with dysvascular transtibial amputees, similarly found lower ground reaction forces with the Flex-Foot foot in comparison with four other feet. Both studies support the hypothesis that higher intact limb forces may be responsible for early degenerative changes, and both show these forces can be modulated by prosthetic foot type of known qualitative differences in stiffness.

Recognizing that the design and properties of the prosthetic foot have a strong influence on performance, Pitkin (28) set out to design a foot that would closely approximate the performance of the biological foot. Pitkin hypothesized that the high initial stiffness in current transtibial limbs has two negative consequences. First, it decreases the range of motion in the knee and thus limits the knee's ability to function as a shock absorber. The second negative consequence is that a stiff ankle can increase the pressure on the residual limb within the socket. Both negative consequences can result in pain, discomfort, and greater risk of ulceration at the residual-limb-socket interface. Pitkin's rolling-joint prosthetic foot was designed to mitigate the effects of high initial stiffness. In testing the prototype, Pitkin and coworkers (32) reported that they observed that their test subject had greater stance phase knee flexion while wearing the prototype in comparison to the knee flexion observed while wearing a SACH foot.

Effect of Prosthetic Feet on Amputee Gait—Energetic Effects

Transtibial amputees are known to expend greater amounts of energy while walking than nonamputees (33). The magnitude of disparity appears to be dependent on the cause of amputation. Dysvascular amputees, walking at a self-selected speed of 45 m/min, demonstrated a cost of locomotion of 0.26 ml/kg/m, while traumatic amputees walked at a faster self-selected speed (71 m/min) and at a lower cost of locomotion (0.20 ml/kg/m). However, neither could compete with the nonamputee control group whose self-selected speed was 82 m/min at a locomotion cost of 0.16 ml/kg/m.

Several investigators have hypothesized that differences between prosthetic limbs might have an effect on

locomotion costs and that certain manufacturers' products might reduce metabolic expenditures. Casillas and coworkers (34) compared the metabolic performance of both traumatic and dysvascular transtibial amputees walking with a SACH foot and an energy-storing foot (Proteor). The traumatic amputees walked 6 percent faster with an 8-percent lower cost of locomotion on the energy-storing foot compared to the SACH foot. However, the dysvascular amputees showed no significant differences between feet. Colborne et al. (35) also looked at metabolic measures, but in children and adolescent congenital amputees walking on either a SACH or Seattle Foot. The cost of locomotion was slightly lower with the Seattle foot, but for only a portion of the 8-minute protocol (statistically significant at minutes 3 and 7). These two studies support the hypothesis that limb properties can affect metabolic costs.

In contrast, several other investigators have rejected the hypothesis that differences between prosthetic limbs would be reflected in metabolic costs. Lehmann and others (21) reported metabolic measures from a mixed group of traumatic and dysvascular transtibial amputees in a study comparing the SACH foot with a Flex-Foot foot and a Seattle Foot. The test subjects walked at various velocities (73 to 120 m/min), but no significant differences in the cost of locomotion were measured. In a related work comparing the SACH foot with a Seattle Ankle/Lite foot with another mixed amputee group (N=10), Lehmann and coworkers came to the same conclusions (20). Also finding no metabolic differences in a mixed group were Perry and Shanfield (23) in a comparison of five different feet during walking. However, both Lehmann's and Perry's metabolic results may be confounded by failing to block effects related to the cause of amputation (traumatic *versus* dysvascular). By using a mixed subject pool with known differences shown by the Waters et al. study (33), the increased variability in the data would require a large sample size in order to detect a significant difference (36).

In another study by Perry's and coworkers (22), they also found no metabolic differences between five different feet during transtibial amputee walking. This study likewise included a mixed pool (traumatic and dysvascular) for analysis, giving cause for potentially confounded results. Perhaps of equal relevance, the investigators also did not alter the prosthetic alignment between successive feet in an attempt to control as many variables as possible. As noted by Michael (37), this failure to optimize alignment between successive feet might inadvertently mask differences between feet, and furthermore, contradicts contemporary standards for prosthetic prescription.

In summary, when metabolic measures of the cost of locomotion were included in the investigation, the results showed small differences or were inconclusive. However, the inconclusive results may be due to sample population problems (mixed traumatic and dysvascular groups) or methodological limitations (alignment issues). The question whether limb properties can affect metabolic measures is unresolved. Differences in limb properties can alter gait biomechanics, and this may reduce dependency on residual and secondary muscles. In turn, the reduced dependency on these less-effective muscles may reduce the cost of locomotion as well as secondary injuries such as lower-back pain, intact limb articular cartilage damage, and soft tissue ulcers, all of which are difficult to measure.

SUMMARY

Stiffness differences between feet have well-recognized effects on gait biomechanics, but somewhat inconclusive effects on metabolic measures. However, in spite of these observations, it remains completely unknown what the optimal stiffness profile should be in order to minimize the metabolic cost of locomotion, increase gait symmetry, or simply improve patient comfort. For clinicians using the current prescription model based on manufacturer guidelines and a qualitative knowledge base differentiating products from various manufacturers, more detailed information is desirable.

It remains a long-term aim to explore how stiffness should vary in the prosthetic limbs of transtibial amputees as they go about their activities in their daily lives. By learning what stiffness profile is appropriate and how it should change with respect to variation in walking speed and daily activities, more effective prosthetic limbs that further vocational and recreational pursuits can be developed.

DISCUSSION

The recent data provided by Legro et al. (1) informs us that the key issues of importance to the lower-limb amputee are residual-limb comfort and optimum biomechanical restoration. In light of this, it is troubling to note the results of Ehde's (3) detailed survey of post-amputa-

tion pain in a large number of lower extremity amputees: Despite recent innovations, prosthetic replacement still results in a high incidence of moderate to severe residual-limb pain. In the absence of data on their current prescription, it is uncertain whether the pain experienced by the amputees in the survey could be modulated by VSAP or by changes in the stiffness characteristics of their prosthetic feet. However, several studies suggest that this might be the case (14,16).

High-frequency impact loads, or transients, occur primarily within the first 20 to 30 msec of heel contact. These high-frequency transients are felt to be particularly injurious to soft tissues and other connective tissue structures. The absorption of these transients is critically related to three characteristics of the prosthesis: the heel stiffness, the characteristics of a vertical shock pylon if present, and the characteristics of the interface material. From a clinical perspective, particular attention must be made to the prosthetic prescription when the residual limb is relatively short and therefore has a reduced surface area for force dissipation; the activity level is high (i.e., sports participation) such that the load magnitude and/or the number of cycles is high; or the soft tissues of the residual limb are compromised. In these instances, the decision whether to utilize special interface materials or shock-absorbing pylons and the choice of heel stiffness characteristics become particularly important. These determinations are currently based upon clinical experience and marketing information. There is little objective information that can assist the clinician in intelligent decision making as to optimum stiffness characteristics of prosthetic components or which VSAP have the greatest efficacy under specific loading conditions. There is a critical need for further study of prosthetic components and their effect on patient function.

The ideal prosthetic prescription would restore the biomechanical characteristics of the intact lower limb. In the transtibial amputee, this would include the normal energy dissipative, storage, and generative functions of the foot-ankle musculotendinous structures. As noted earlier, current prosthetic components are not able to generate energy nor are they able to adapt to various walking speeds or surface conditions. Some do allow for passive energy storage and return, but these are fabricated with a single-stiffness profile that cannot function optimally across a spectrum of activity levels and functional demands. Presently, there are no biomechanical data or metabolic energy consumption data that conclusively show one prosthetic foot type to have a measurable benefit over another. Clinical decision-making is therefore largely empirical, based upon cost analyses and experience. Enhancements in function will largely come from a better understanding of the effects of prosthetic foot stiffness profiles on the biomechanical function of the amputee, and the development of intelligent foot systems that may allow dynamic alterations in foot stiffness based upon specific loading characteristics.

REFERENCES

1. Legro MW, Reiber G, del Aguila M, Ajax MJ, Boone DA, Larsen JA, et al. Issues of importance reported by persons with lower limb amputations and prostheses. J Rehabil Res Dev 1999;36(3):155–63.
2. Postema K, Hermens HJ, de Vries J, Koopman HF, Eisma WH. Energy storage and release of prosthetic feet. Part 2: Subjective ratings of 2 energy storing and 2 conventional feet, user choice of foot and deciding factor. Prosthet Orthot Int 1997;21(1):28–34.
3. Ehde D, Czerniecki JM , Smith DG, Campbell KM, Edwards WT, Jensen MP, Robinson LR. Chronic phantom sensations and pain following lower limb amputation. Arch Phys Med Rehabil 2000;81(8):1039–44.
4. Kegel B, Webster JC, Burgess EM. Recreational activities of lower extremity amputees: a survey. Arch Phys Med Rehabil 1980;61(6):258–64.
5. Antonsson EK, Mann RW. The frequency content of gait. J Biomech 1985;18:39–47.
6. Serink MT, Nachemson A, Hansson G. The effect of impact loading on rabbit knee joints. Acta Orthop Scand 1977;48(3):250–62.
7. Radin EL, Ehrlich MG, Chernack R, Abernethy P, Paul IL, Rose RM. Effect of repetitive impulsive loading on the knee joints of rabbits. Clin Orthop 1978;131:288–93.
8. Radin EL, RB Orr, L Kelman, IL Paul, Rose RM. Effect of prolonged walking on concrete on the knees of sheep. J Biomech 1982;15:487–92.
9. Voloshin A, Wosk J. An *in vivo* study of low back pain and shock absorption in the human locomotor system. J Biomech 1982;15(1):21–7.
10. Sandover J. Dynamic loading as a possible source of low-back disorders. Spine 1983;8(6):652–8.
11. Boulton AJM. The importance of abnormal foot pressures and gait in the causation of foot ulcers. In: Connor H, Boulton AJM, Ward JD, eds. The foot in diabetes. Chichester: Wiley. 1987; p. 11–21.
12. Boulton AJM. The diabetic foot. Med Clin North Amer 1988;72:1512–30.
13. Goldstein B, Sanders J. Skin response to repetitive mechanical stress: a new experimental model in pig. Arch Phys Med Rehabil 1998;79(3):265–72.

14. Miller LA, Childress DS. Analysis of a vertical compliance prosthetic foot. J Rehabil Res Dev 1997;34(1):52–7.
15. Gard S. Mechanical characterization of vertical shock-absorbing pylons for lower limb prostheses. Proceedings of the North American Congress on Biomechanics (NACOB 98); 1998 August 14–18, Waterloo, Ontario, Canada.
16. Hsu MJ, Nielsen DH, Yack HJ, Shurr DG. Physiological measurements of walking and running in people with transtibial amputations with 3 different prostheses. J Orthop Sports Phys Ther 1999;29(9):526–33.
17. Gitter A, Czerniecki JM, DeGroot DM. Biomechanical analysis of the influence of prosthetic feet on below-knee amputee walking. Am J Phys Med Rehabil 1991;70(3):142–8.
18. Ehara Y, Beppu M, Nomura S, Kunimi Y, Takahashi S. Energy storing property of so-called energy storing prosthetic feet. Arch Phys Med Rehabil 1993;74(1):68–72.
19. Menard MR, McBride ME, Sanderson DJ, Murray DD. Comparative biomechanical analysis of energy storing prosthetic feet. Arch Phys Med Rehabil 1992;73(5):451–8.
20. Lehmann JF, Price R, Boswell-Bessette S, Dralle A, Questad K. Comprehensive analysis of dynamic elastic response feet: Seattle Ankle/Lite Foot *versus* standard SACH Foot. Arch Phys Med Rehabil 1993a;74:853–61.
21. Lehmann JF, Price R, Boswell-Bessette S, Dralle A, Questad K, deLateur BJ. Comprehensive analysis of energy storing prosthetic feet: Flex-Foot and Seattle Foot *versus* standard SACH Foot. Arch Phys Med Rehabil 1993b;74:1225–31.
22. Torburn L, Perry J, Ayyappa E, Shanfield SL. Below-knee amputee gait with dynamic elastic response prosthetic feet: a pilot study. J Rehabil Res Dev 1990;27(4):369–84.
23. Perry J, Shanfield S. Efficiency of dynamic elastic response prosthetic feet. J Rehabil Res Dev 1993;30(1):137–43.
24. van Jaarsveld HW, Grootenboer HJ, de Vries J, Koopman HF. Stiffness and hysteresis properties of some prosthetic feet. Prosthet Orthot Int 1990;14(3):117–24.
25. Cortes A, Viosca E, Hoyos JV, Prat J, Sanchez-Lacuesta J. Optimisation of the prescription for trans tibial (TT) amputees. Prosthet Orthot Int 1997;21(3):168–74.
26. Perry J, Boyd LA, Rao SS, Mulroy SJ. Prosthetic weight acceptance mechanics in transtibial amputees wearing the Single Axis, Seattle Lite, and Flex Foot. IEEE Trans Rehabil Eng 1997;5(4):283–9.
27. Saunders JBdeCM, Inman VT, Eberhart HD. The major determinants in normal and pathological gait. J Bone Joint Surg 1953;35A(3):543–57.
28. Pitkin MR. Synthesis of a cycloidal mechanism of the prosthetic ankle. Prosthet Orthot Int 1996;20(3):159–71.
29. Torburn L, Schweiger GP, Perry J, Powers CM. Below-knee amputee gait in stair ambulation. A comparison of stride characteristics using five different prosthetic feet. Clin Orthop 1994;303:185–92.
30. Powers CM, Torburn L, Perry J, Ayyappa E. Influence of prosthetic foot design on sound limb loading in adults with unilateral below-knee amputations. Arch Phys Med Rehabil 1994;75(7):825–9.
31. Snyder RD, Powers CM, Fontaine C, Perry J. The effect of five prosthetic feet on the gait and loading of the sound limb in dysvascular below-knee amputees. J Rehabil Res Dev 1995;32(4):309–15.
32. Quesada PM, Pitkin M, Colvin J. Biomechanical evaluation of a prototype foot/ankle prosthesis. IEEE Trans Rehabil Eng 2000;8(1):156–9.
33. Waters RL, Perry J, Antonelli D, Hislop H. Energy cost of walking amputees: the influence of level of amputation. J Bone Joint Surg Am 1976;58(1):42–6.
34. Casillas JM, Dulieu V, Cohen M, Marcer I, Didier JP. Bioenergetic comparison of a new energy-storing foot and SACH foot in traumatic below-knee vascular amputations. Arch Phys Med Rehabil 1995;76(1):39–44.
35. Colborne GR, Naumann S, Longmuir PE, Berbrayer D. Analysis of mechanical and metabolic factors in the gait of congenital below knee amputees. A comparison of the SACH and Seattle feet. Am J Phys Med Rehabil 1992;71(5):272–8.
36. Czerniecki, JM, Gitter A. Prosthetic feet: A scientific and clinical review of current components. Phys Med Rehabil 1994;8(1):109–28.
37. Michael JW. Re: Below-knee amputee gait with dynamic elastic response prosthetic feet: a pilot study. J Rehabil Res Dev 1992;29(1):x.