Prosthetics and Orthotics International, 1994, 18, 180-190

Technical note

A pilot study to test the influence of specific prosthetic features in preventing trans-tibial amputees from walking like able-bodied subjects

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Abstract

The purpose of this pilot investigation was to develop a method to test the influence of specific prosthetic features in preventing trans-tibial amputees from walking like able-bodied subjects. An able-bodied subject was fitted with a patellar-tendon-bearing orthosis incorporating several features of an amputee's prosthesis. Kinetic, kinematic and metabolic data were collected as features were systematically removed from the orthosis. While wearing the orthosis the gait of the able-bodied subject closely simulated trans-tibial amputee gait kinematically, kinetically and metabolically. Although it was obvious that the various prosthetic features influenced the kinetics and kinematics of gait, they were difficult to quantify with only a single subject. However, the two features which appeared to have the largest influence in preventing trans-tibial amputees from walking like able-bodied subjects were patellar tendon loading and a solid ankle.

Introduction

It has been well documented that trans-tibial amputees (TTAs) do not walk like able-bodied individuals (Breakey, 1976; Doane and Holt, 1983; Culham *et al.*, 1984; Lewallen *et al.*, 1986; Winter and Sienko, 1988; Smith, 1990). Engsberg *et al.* (1993) stated that TTAs will now walk like able-bodied (AB) subjects until a prosthesis is developed that functions like an intact leg and foot. If the design of such a device is to be undertaken it is critical to understand the relative contribution of each component of the prosthesis towards permitting or preventing TTAs from walking like ABs.

A patellar-tendon-bearing prosthesis with a Symes foot terminal device includes the following major components: a) a solid ankle allowing no dorsiflexion-plantarflexion or eversion-inversion, b) substantial loading on the patellar ligament and other soft tissue regions of the stump, c) passive flexion-extension at the metatarsal-phalangeal joints, d) slight knee flexion imposed by the socket and thus preventing full knee extension, and e) a cushioned heel to assist the foot in attaining a foot-flat position. The contribution of each of these components towards permitting or preventing TTAs from walking like ABs is presently unknown. For example, the lack of dorsiflexion-plantarflexion in the solid ankle of the foot must prevent normal walking. It would seem that the extent of this prevention should be quantifiable.

Two difficulties arise when considering the use of an amputee as a subject. The first is that adding and removing components of a prosthesis is impossible since the amputee requires the prosthesis to function. The second is that if components were removed, they could not be replaced with able-bodied functions, thus, it would be impossible to directly determine the influence of the component in preventing the TTA from walking like an AB. However, if an orthosis could be created for an AB that functions like a prosthesis, features of the prosthesis could be systematically varied to

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help understand the numerical contribution of each component. The purpose of this investigation was to develop a method to test the influence of specific prosthetic features in preventing TTAs from walking like ABs.

Methods

An able-bodied individual fitted with an ankle-foot orthosis (AFO) that incorporated the main features found in an amputee's prosthesis was used as a subject. The features were systematically removed one at a time and replaced with actual able-bodied functions throughout the protocol. As a result each consecutive orthosis became increasingly more like an AB limb and less and less like an amputee's prosthesis. By using this protocol, the relative differences of the observed variables between the AFOs could be used to determine the influences of specific prosthetic features on gait.

Subject

The AB subject who volunteered for this study was a 24 year old, 165 cm tall male of 59.15 kg mass. It should be noted that throughout the document the legs of the AB subject are referred to as prosthetic and non-prosthetic. While this is not the exact subject's condition, it clearly describes the limitations that were imposed upon him.

Preliminary investigations

The first prototype AFO, constructed by a registered orthotist, had a Kingsley Symes (size 7) foot attached to its bottom, thus incorporating the cushioned heel and forefoot features of the prosthetic foot. With the foot attached, a lift of approximately 7 cm had to be fabricated in the shoe of the non-prosthetic leg to compensate for the height difference. Although preliminary tests showed that this lift height did not significantly affect the subject's kinetics, the kinematics of his gait were altered. Thus, it was necessary to reduce the lift height associated with the non-prosthetic leg. To decrease the lift height, the most obvious solution was to remove the Symes foot from the bottom of the AFO and incorporate the characteristics of the foot directly into the AFO.

Load-deformation tests were performed with an Instron materials testing machine on the Symes heel and forefoot to determine to what extent the parts of the foot were influenced by the forces associated with walking for the AB subject.

To test the deformation of the Symes heel, the Symes foot was placed in the testing machine at a number of different angles representing relevant angles found between the

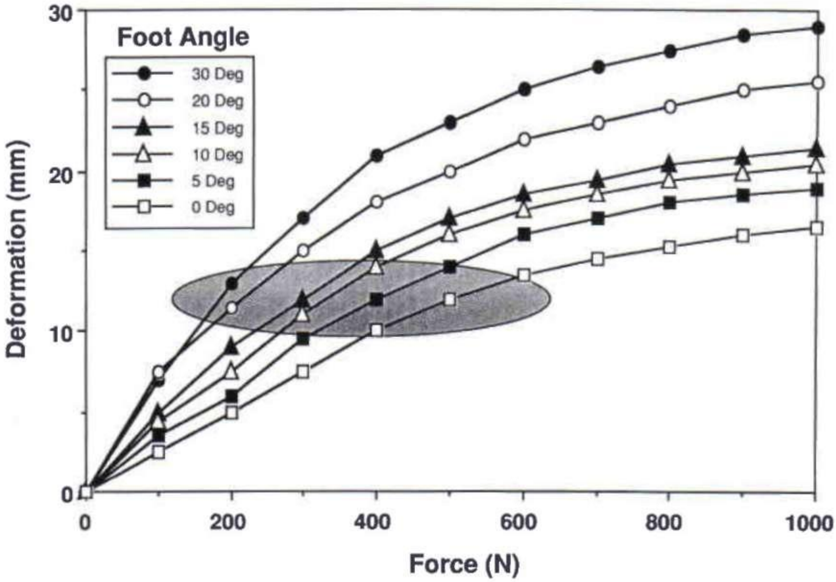

Fig. 1. Symes heel deformation curves at various foot angles. Shaded region indicates deformations for foot angles corresponding to forces produced by the AB subject.

sole of the foot and the floor while walking (0°-30°). At each of the different angles the heel was compressed from 0-1000 N (0-1.7 BW) and the deformation was recorded (Fig. 1). The vertical forces produced by the AB subject during walking that corresponded to these foot angles lie in the shaded region of the graph. The maximum deformation was about 15 mm which correlated to a change in foot angle of approximately 3°. Since the main purpose of the cushioned heel is to hasten the foot to a full ground contact position and since the change in foot angle was minimal, it was decided to eliminate the cushioned heel from the pilot study. This permitted a significant reduction in the lift height required for the shoe on the non-prosthetic leg.

The deformation of the Symes forefoot was tested at various distances from the heel by applying a normal load along a thin section (approximately 6 mm) of the plantar surface of the forefoot (Fig. 2). The forefoot was loaded until the limitations of the measurement methods were reached (i.e. until the deformation of the forefoot caused loads to no longer be applied perpendicular to the plantar surface of the foot resulting in slippage). The solid lines represent actual data collected while the dashed lines are linear extrapolations. By correlating the magnitude of the vertical ground reaction force produced by the subject while walking to the appropriate point of application, the approximate deformation of the forefoot was determined (shaded region of Figure 2). Since the deformation of the forefoot was substantial (maximum of approximately 40 mm) it was decided that passive forefoot flexion-extension was an important feature of the Symes foot and should remain a part of the investigation.

As a result of the preliminary deformation tests the orthotist fabricated a new AFO for the protocol without an actual prosthetic foot. The new AFO had the following features: a) a solid ankle, b) patellar tendon loading, c) passive flexion-extension of the forefoot, and d) slight knee flexion, however, it did not include a cushioned heel. The forefoot deformation was incorporated directly into the base of the AFO by thinning the plastic and placing the toe-break at a location similar to that of the Symes foot.

A final preliminary measure was to ensure the subject was non-weight bearing at the heel (i.e. the load was being supported by the patellar tendon and other tissues). A Tekscan pressure insole was placed underneath the sole of the subject's foot and several standing and walking trials were collected. In all trials the

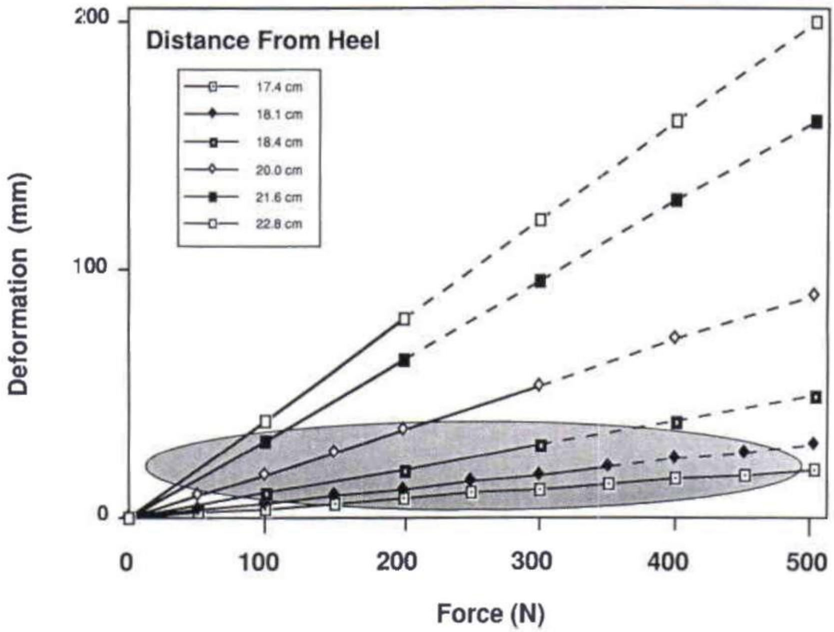

Fig. 2. Symes forefoot deformation curves at various distances from the heel. Shaded region indicates deformations for distances from the heel corresponding to forces produced by the AB subject.

pressure at the heel was below the threshold of the Tekscan insole (40 kPa) thus ensuring that the weight of the subject was being supported by the patellar tendon and other weight bearing regions.

Protocol

The study began with the collection of kinematic and kinetic data on the AB subject while walking normally without any AFO. The subject was then fitted with the first ankle-foot orthosis (AFO1) incorporating: a) a solid ankle, b) patellar-tendon-bearing, c) passive flexion-extension of the forefoot at the metatarsal-phalangeal joints, and d) slight knee flexion (approximately 10°). Kinetic and kinematic data were collected immediately after the subject was fitted with AFO1 (i.e. AFO1-I). The subject then underwent a two week period where he wore AFO1 for 2-3 hours a day and walked in AFO1 a minimum of 30 minutes/day. During this period a gait training programme with a registered physiotherapist who regularly trained trans-tibial amputees was undertaken. At the end of the two week wearing period, the subject again had kinematic and kinetic data collected (i.e. AFO1-F). Oxygen uptake data while walking normally and walking with AFO1 were also collected. The initial and final data collection sessions (i.e. AFO1-I and AFO1-F) were designed to evaluate any adaptation that occurred over the wearing period.

AFO1 was then modified by removing knee flexion resulting in 0° flexion in the orthosis. This second ankle-foot orthosis (AFO2) now had the features of: a) a solid ankle, b) patellar-tendon-bearing, and c) passive flexion-extension of the forefoot. Again kinematic and kinetic data were collected upon the initial fitting of AFO2 (i.e. AFO2-I). Having completed the physiotherapy treatment the subject underwent a one week wearing period of 2-3 hours a day ensuring that he walked in AFO2 a minimum of 30 minutes per day. Final fit kinetic and kinematic data were collected after the one week wearing period (i.e. AFO2-F).

Next AFO2 was modified by removing the passive flexion-extension of the forefoot allowing the subject to actively flex and extend his forefoot. The remaining features of the third ankle-foot orthosis (AFO3) were a) a solid ankle, and b) patellar-tendon-bearing. As with the protocol for AFO2, initial fit data were collected at the first fitting and final fit data were collected after a one week wearing period for AFO3.

The final modification was accomplished by removing the patellar-tendon-bearing feature from AFO3 and allowing the subject to bear weight on his foot. Hence, the fourth ankle-foot orthosis (AFO4) retained only a solid ankle preventing any dorsiflexion-plantarflexion or eversion-inversion of the ankle joint. Again initial fit data for AFO4 were collected and final fit data were collected one week later after wearing.

One week after final data collection with AFO4, normal kinetic and kinematic data (without an AFO) were again collected. The data from this second normal session were averaged with data from the first normal session to provide a composite of normal data.

By using this protocol, the relative differences between the AFOs could be used to determine the influences of specific prosthetic features on gait. For example, the difference between the normal data and the data for AFO4 would be attributed solely to the solid ankle and the difference between the data collected for AFO4 and AFO3 would be due to patellar tendon loading. Table 1 summarises the different features incorporated in each AFO.

Kinetics

The methods used to collect and evaluate kinetic data were similar to those previously reported by Engsberg *et al.* (1993). Briefly, vertical, anteroposterior, and mediolateral force data were collected using two adjacent force

Table 1. The various prosthetic features incorporated in each ankle-foot orthosis

Ankle-Foot Orthosis (AFO)	Features
AFO1	solid ankle patellar-tendon-bearing passive flexion-extension of the forefoot slight knee flexion
AFO2	solid ankle patellar-tendon-bearing passive flexion-extension of the forefoot
AFO3	solid ankle patellar-tendon-bearing
AFO4	solid ankle

platforms sampling at 1000 Hz. Data from six trials were collected for each condition in which three trials were collected with the prosthetic (left) leg landing first and three trials were collected with the non-prosthetic (right) leg landing first. The freely chosen walking speed determined from preliminary data collection sessions (1.6 m/s ± 10%) was enforced for each trial.

Force data were normalised by dividing by body weight and the total time spent of both plates was normalised to a value of 1. Average force-time curves of the six trials for each condition were determined. From the force-time curves, discrete variables were obtained for use in comparison. The discrete variables used in this investigation included first local vertical (ZMax1), second local vertical (ZMax2), anterior and posterior maxima (RMax and PMax, respectively).

Kinematics

Kinematic data were collected simultaneously with the kinetic data using a four camera video system sampling at 200 Hz. Three reflective spheres of 1 cm diameter were placed on the foot, shank, thigh and trunk to permit 3-dimensional analysis.

The coordinate systems used to determine angular orientations of the body segments were the same as those used by Engsberg *et al.* (1992). Consistent placement of the markers allowed relative comparisons between conditions, however, absolute angles had little or no meaning as joint centres were not determined for this pilot study.

Three trials were analysed for each condition and average angular position-time curves of the trials were determined. From the angular position-time curves, discrete variables were obtained for use in comparison. The discrete variables included sagittal plane touchdown angles of the trunk and thigh for both the prosthetic and non-prosthetic limbs (Engsberg *et al.*, 1992).

Metabolic

The methods used to collect metabolic data were similar to those used by Herbert *et al.* (in press). Briefly a Quinton Instruments Model 24-72 Treadmill System (Quinton Instruments, 2121 Terry Avenue, Seattle, Washington 98121) and Horizon Metabolic Measurement Cart System (SensorMedics Corporation, 1630 South State College Boulevard, Anaheim, California 92806) were used to measure oxygen uptake in millilitres/minute (VO_2). Heart rate in beats/minute (HR) was continuously monitored using a Polar Sport Tester PE 3000 (Polar U.S.A. Incorporated, 470 West Avenue,

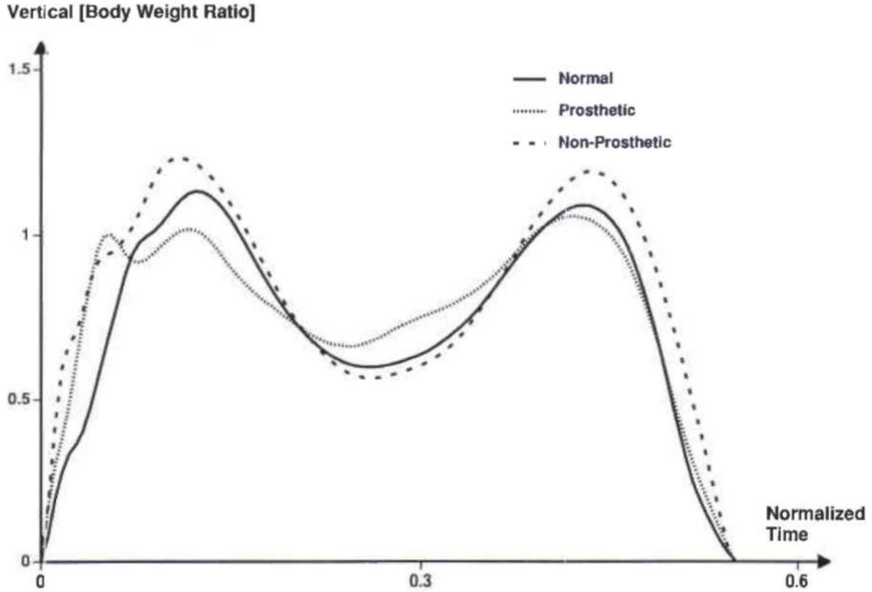

Fig. 3. Average vertical force-time curve for the subject walking normally and for the prosthetic and non-prosthetic legs while wearing AFO1.

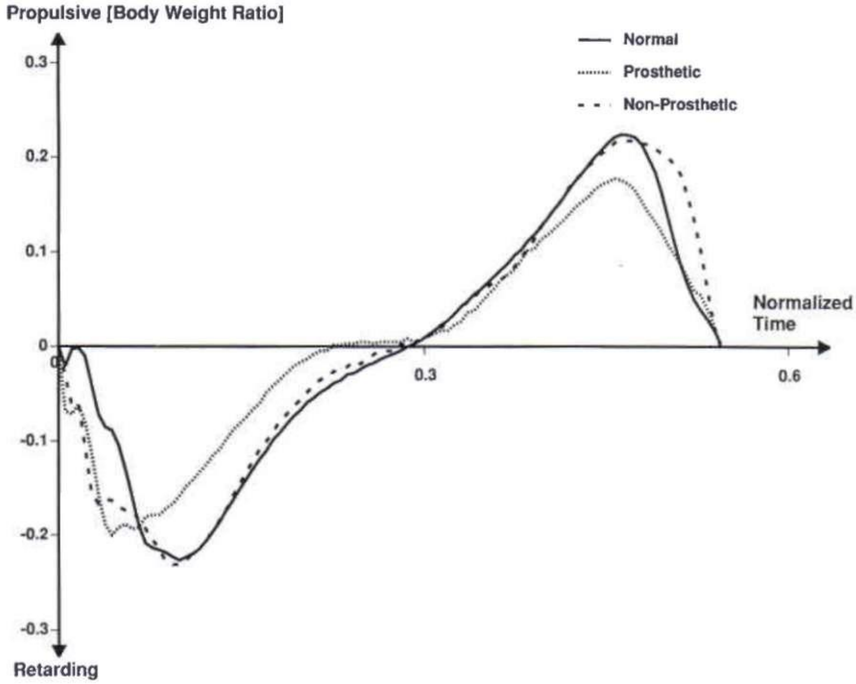

Fig. 4. Average anteroposterior force-time curve for the subject walking normally and for the prosthetic and non-prosthetic legs while wearing AFO1.

Stanford, Connecticut 109602) secured around the chest.

The subject was fitted with a noseclip and a headgear apparatus securing the mouthpiece to the expired gas hose. The subject then walked on the treadmill for 2 minutes at each of four speeds (freely chosen walking speed (1.6 m/s), 20% above freely chosen walking speed, 20% below freely chosen walking speed and a fixed speed of 1.2 m/s) in a randomly assigned order. Heart rate was recorded every 30 seconds throughout the test while oxygen consumption (ml/(kg·min)) was taken at 2 minute intervals. Oxygen consumption relative to distance (ml kg-m)) (SVO_2) was then calculated.

Results

The data presented will be from the final fittings of each AFO only (i.e. AFO1-F to AFO4-F). A comparison of data collected from the initial fitting to data collected from the final fitting is beyond the scope of this paper.

Figures 3 and 4 show final fit average force-time curves for the subject. Figure 3 shows the vertical force-time curves for the AB subject walking normally and for the non-prosthetic and prosthetic leg of the subject while wearing AFO1 (AFO1-F). Similarly, Figure 4 shows the anteroposterior (retarding-propulsive) force-time curves.

Figures 5, 6, 7 and 8 are results of discrete vertical and anteroposterior variables for both the prosthetic and non-prosthetic legs. Each figure compares the value of the discrete variable while walking normally to the value while walking with each AFO. Included in the comparisons and indicated by an asterisk are data collected by Engsberg *et al.* (1993) on AB and TTA children. For this study, AFO1 corresponds to the TTA case and normal correlates to the AB case. Figures 9 and 10 display final fit results of sagittal plane touchdown angles of the trunk and thigh, respectively, for both the prosthetic and non-prosthetic limbs (AFO1-F to AFO4-F).

Figure 11 compares SVO_2 during normal walking to SVO_2 during walking while wearing AFO1 for various walking speeds. For each walking speed, the oxygen uptake while wearing AFO1 was higher than while walking normally. When all speeds were combined, there was an average increase of 16% SVO_2 while walking with AFO1.

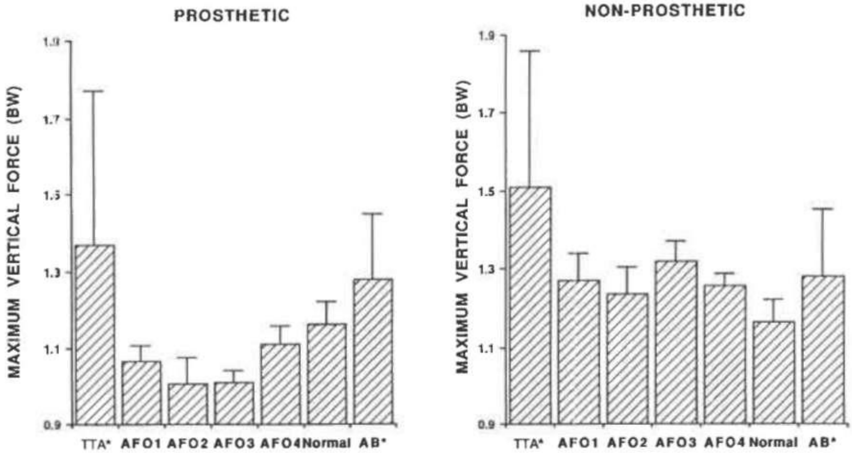

Fig. 5. Prosthetic and non-prosthetic comparisons of the first local maximum (ZMax1) of the vertical force-time curve for walking with the different AFOs and walking normally. *Indicates data included in the comparison for TTA and AB children from Engsberg *et al.* (1993).

Discussion

The purpose of this investigation was to develop a method to test the influence of specific prosthetic features in preventing TTAs from walking like ABs.

Since the investigation was designed only to develop a method and only a single subject was involved, no statistical treatment was performed on the data to determine significant differences between variables. A limitation of the study was that although the methods used to collect kinematic data allowed comparisons amongst conditions within this study (i.e. surface marker locations were consistent for all AFOs), the data from this investigation could not be quantitatively compared with data from previous investigations in the literature. Another limitation was that data were collected only while walking on level ground. Running and/or walking on uneven ground should be included in future investigations as the influence of the various prosthetic features could possibly change.

The three key elements of this investigation were to determine if: 1) AFOs could be fabricated that would incorporate the features of

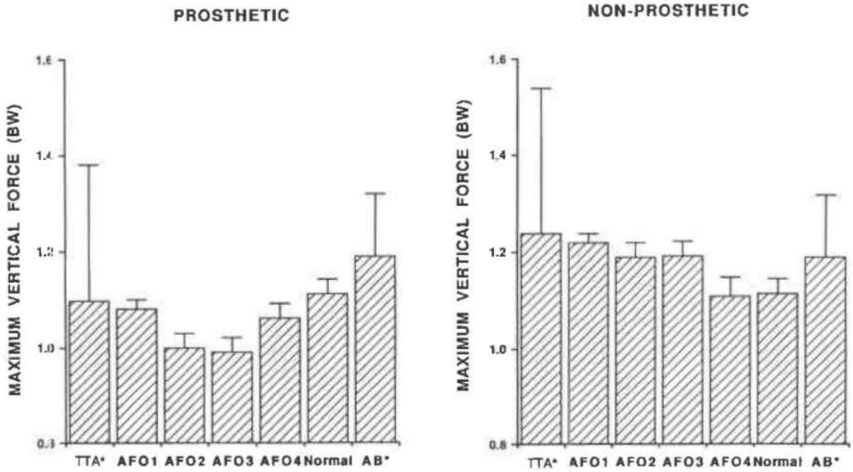

Fig. 6. Prosthetic and non-prosthetic comparisons of the second local maximum (ZMax2) of the vertical force-time curve for walking with the different AFOs and walking normally. *Indicates data included in the comparison for TTA and AB children from Engsberg *et al.* (1993).

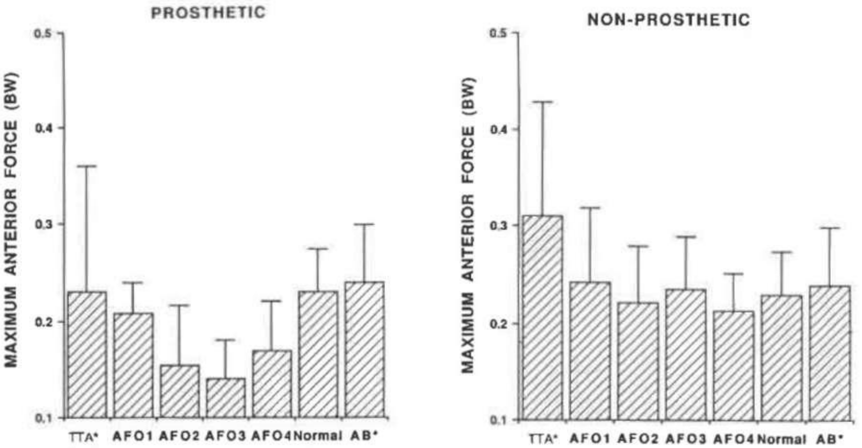

Fig. 7. Prosthetic and non-prosthetic comparisons of the maximum anterior force (RMax) for walking with the different AFOs and walking normally. *Indicates data included in the comparison for TTA and AB children from Engsberg *et al.* (1993).

a Symes foot prosthesis; 2) an AB subject wearing the AFOs would behave similar to a TTA and, 3) detectable differences in gait existed between AFOs with different features.

Regarding the first key element, it was possible to fabricate a single AFO with the following prosthetic features: a) a solid ankle, b) patellar tendon weight bearing, c) passive flexion-extension at the metatarsal phalangeal joints and d) slight knee flexion. It was also possible to systematically modify the AFO to examine the various features. This was an important factor since the cost associated with fabricating many AFOs for a large group of subjects would be quite high.

Information regarding the second key element is obtained by comparing data from the literature collected on TTA and AB subjects to data collected in this study. The average vertical and anteroposterior curves. (Figs. 3 and 4) were similar to those published by Engsberg *et al.* (1993). In both studies the average vertical ground reaction force maxima for the prosthetic leg were lower than normal while the maximum forces for the non-prosthetic leg were higher than normal. Similarly, in both studies, the maximum anteroposterior forces for the prosthetic leg were lower than normal while the

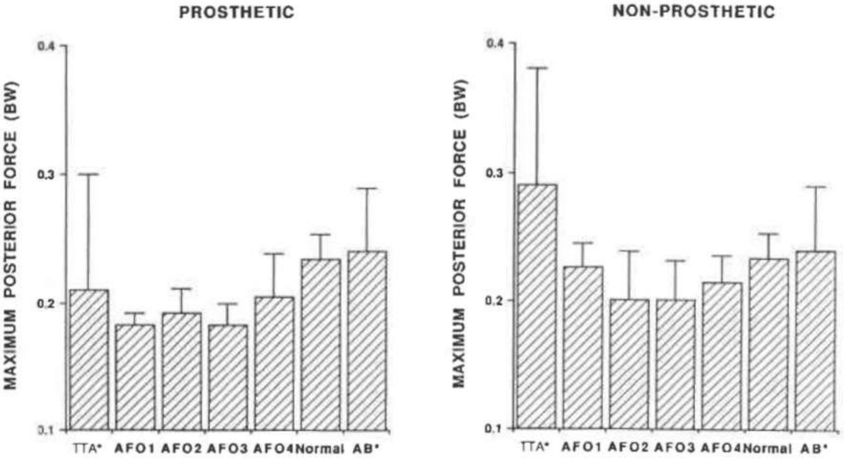

Fig. 8. Prosthetic and non-prosthetic comparisons of the maximum posterior force (PMax) for walking with the different AFOs and walking normally. *Indicates data included in the comparison for TTA and AB children from Engsberg *et al.* (1993).

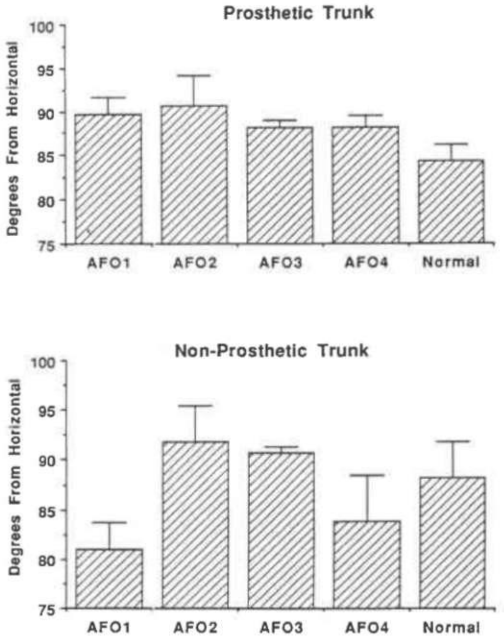

Fig. 9. Prosthetic and non-prosthetic comparisons of the sagittal touchdown angle of the trunk for walking with the different AFOs and walking normally.

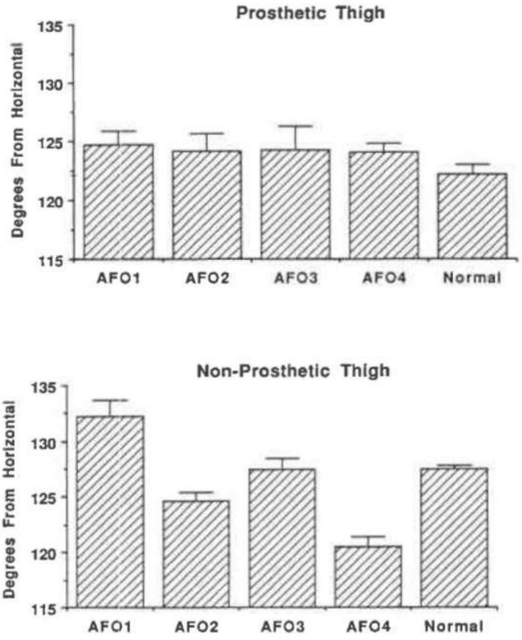

Fig. 10. Prosthetic and non-prosthetic comparisons of the sagittal touchdown angle of the thigh for walking with the different AFOs and walking normally.

forces for the non-prosthetic leg were slightly higher or equal to normal.

Sagittal plane knee angle data collected by Culham *et al.* (1984) on 10 TTA adults and by Engsberg *et al.* (1992) on 3 TTA children found touchdown knee angles to be larger in the non-prosthetic limb than the prosthetic limb. Engsberg *et al.* (1992) also found that AB children had smaller knee angles at touchdown in the sagittal plane than TTA children. The subject in this study had a knee touchdown angle at 10° when walking normally, 14° for the prosthetic limb and 15° for the non-prosthetic limb. As previously mentioned, the quantities cannot be compared directly to those of other studies, however, the trends appear to be similar to those found in actual TTAs.

Engsberg *et al.* (1992) found the trunk segment for TTA children had a greater amount of forward flexion on touchdown than that of AB children. This was the case for both non-prosthetic and prosthetic support. While wearing AFO1 in this study, the subject's trunk also had more forward flexion during non-prosthetic support but less forward flexion during prosthetic support (Fig. 9). Engsberg *et al.* (1992) also found that the thighs of the TTA children were oriented more horizontally than that of the AB children at touchdown. A similar trend was apparent in this study (Fig. 10) where the subject's thigh was oriented more horizontally at touchdown for both legs when walking with AFO1 than when walking normally.

When TTA adults walk at the same speed as AB adults, their energy expenditure is approximately 9% greater (Gonzalez *et al.*, 1974; Ganguli *et al.*, 1974). The metabolic data shown in Figure 11 corresponded to this data as well as data collected by Herbert *et al.* (in press) on TTA and AB children. Herbert *et al.* found SVO_2 to be significantly higher at all speeds of walking for TTA children. When averaging all speeds together, the subject consumed 16% more oxygen when walking with AFO1 than when walking normally.

It appears from the kinetic, kinematic and metabolic data in the literature that the gait of the AB subject while wearing AFO1 closely resembled TTA gait. Similarities existed in oxygen consumption, vertical and antero-posterior ground reaction forces, and sagittal knee and thigh angles. Only slight differences were found in the sagittal trunk angles, however, these differences were not substantial and overall it appeared that TTA gait was successfully simulated with an AB subject wearing an orthosis with various prosthetic features.

Regarding the third key element of this investigation, actual influences of the prosthetic features were difficult to quantify with measurements on only a single subject. However, some trends were apparent. The largest influence of the various prosthetic features on the kinematics of gait appeared to be in the non-prosthetic limbs. The sagittal touchdown angles for the prosthetic trunk and thigh did not appear to change dramatically (less than 5° in all cases). However, the sagittal touchdown angles for the non-prosthetic trunk and thigh seemed to be influenced to a much larger degree although no obvious pattern developed.

In comparing discrete kinetic variables (Figs. 5, 6, 7, and 8) it was assumed that the prosthetic and non-prosthetic limbs should have values equal or at least very similar to normal if a TTA is to walk like an AB. Thus, the hypothesis was that any prosthetic feature which caused the variable to move toward normal was a positive feature and any feature which caused the variable to move away from normal was a negative feature.

The first and second local vertical maxima, the anterior maximum and to a certain extent the posterior maximum followed a similar trend for the prosthetic limb. The removal of slight knee flexion (AFO2) created a negative effect, the removal of passive forefoot flexion-extension (AFO3) appeared to have little or no effect, the removal or patellar-tendon-bearing (AFO4) had a positive effect and since the values for AFO4 were less than normal in all cases, the removal of the solid ankle would have a positive effect. In each of these comparisons the values associated with the AFO1 and normal situations corresponded reasonably well with respective values for TTA and AB children (Engsberg *et al,* 1993).

A key qualitative result when considering the influence of the various prosthetic features was the subjective rating of importance of the prosthetic features in preventing the subject from walking normally. He rated the relative importance of the features in preventing TTAs from walking like ABs as follows (from most (1) to least (4) important): 1) patellar-tendon-bearing, 2) solid ankle, 3) passive flexion-extension of the forefoot, 4) slight knee flexion. The subject felt the ability to bear weight on the foot was the single most important feature in allowing him to walk more like an AB. Another feature which had a large influence was the solid ankle and the subject felt this feature was of great importance when walking on uneven

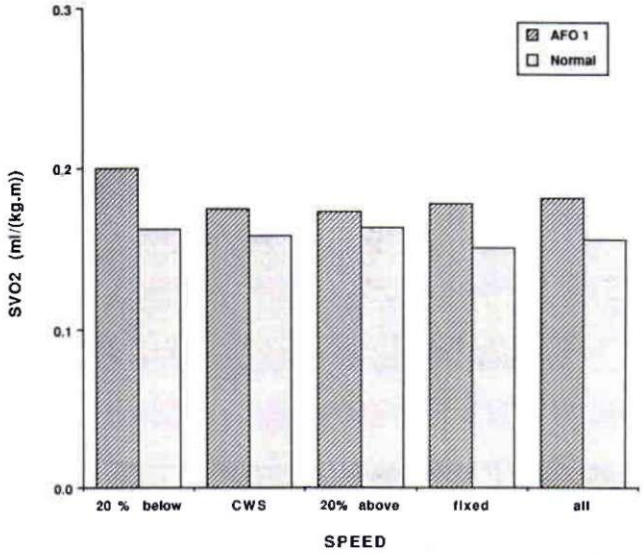

Fig. 11. Oxygen consumption, relative to speed, results for subject walking normally and while wearing AFO1. CWS is the chosen walking speed, 20% below is 20% below CWS, 20% above is 20% above CWS, fixed speed is 1.2 m/s, and all represents a summation of all speeds.

ground. The removal of both knee flexion and passive flexion-extension of the forefoot had small influences on gait according to the subject.

From the kinetic data and the supporting subjective rating it appeared that the patellar-tendon-bearing and the solid ankle had the largest negative influence. The flexion-extension of the forefoot did not appear to have influenced gait during walking. Also, it appeared the slight knee flexion was a positive feature incorporated in the prosthesis to compensate somewhat for the solid ankle and patellar-tendon-bearing. These results are important since current research is directed solely at developing better feet to improve function of the TTA. This data indicates that the loading of the stump may be at least as important. Research directed at alternative loading scenarios may be necessary if TTAs are to walk more like ABs.

In future studies it will be important to include a large number of subjects to see if similar trends appear and to be able to quantify the influences of the various prosthetic features. It will also be important to include other activities in the protocol to determine the influence of the prosthetic features during activities other than level walking (i.e. walking on uneven ground and running.)

Acknowledgments

The authors would like to thank Mark Schneider for fabrication of the various ankle-foot orthoses and the Hospital for Sick Children Foundation for the financial support (#XG91-078).

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