Prosthetics and Orthotics International, 1993, 17, 27–37

Gait patterns of elderly men with trans-tibial amputations

E. D. LEMAIRE*, F. R. FISHER* and D. G. E. ROBERTSON**

*The Rehabilitation Centre, Ottawa, Ontario, Canada

**University of Ottawa, School of Human Kinetics, Ottawa, Ontario, Canada

Abstract

Gait patterns for the non-amputated leg of eight elderly men with trans-tibial amputations were assessed using kinematic and kinetic measures. Kinematically, the subject's walking speed was faster than expected but less than normative non-amputee data. The stride length was also less than non-amputee norms. Net joint moment and power analyses showed various discrepancies between the amputee subjects and non-amputees. The amputees required a concentric ankle dorsiflexor moment just after heel-strike to help move the lower leg into mid-stance position. The concentric plantarflexor moment at push-off was much larger than comparative data. A large eccentric flexor moment was also found at the hip during late mid-stance. Most of these discrepancies could be explained by the lack of an ankle moment generator on the amputated side of the body.

Introduction

With the "baby boom" generation entering middle age, health care systems will soon be expected to service a large senior citizen population. This may lead to an increase in the number of elderly amputees since non-traumatic loss of limb is most prevalent in the aged (Hunter and Waddell, 1976). Although a satisfactory level of clinical experience with seniors exists in the prosthetics field. Scientific research involving the gait of this group is lacking.

Most of the current gait research has been performed on young men with trans-tibial amputations. Of these studies, the majority involve kinematic stride evaluation on young to middle aged subjects (Breakey, 1976; Doane and Holt, 1983; Enoka *et al.*, 1982; Ganguli *et al.*, 1974; Gonzalez *et al.*, 1974; Hannah, *et al.*, 1984; Robinson *et al.*, 1977). These studies describe a population which exhibits an asymmetric gait pattern and walks slower than non-amputees.

Studies involving the kinetics of trans-tibial amputee gait have provided important information regarding amputee locomotion. Seliktar and Mizrahi (1986) used force plate analysis to develop a clinical technique providing quantitative measures for prosthesis alignment. It was suggested that the vertical impulse ratio (ratio between the vertical impulse for the prosthetic leg and the vertical impulse for the sound leg), antero-posterior impulse ratio, and perturbations on the antero-posterior force curve would be adequate for the assessment of prosthetic alignment; however, the effects of alignment changes on gait dynamics were found to be transient until the patient had reached a new steady state. The time required for the amputee's gait to stabilize was considered detrimental to the use of force plate analysis in a clinical setting.

Lewallen *et al.* (1986) used net joint moments to examine the load exerted on the joints of children with trans-tibial amputations. These children encountered larger ground reaction forces on their intact leg than non-amputees; however, joint moments at the knee and the hip were less than or equal to results for normal children (even though the ankle produced a greater dorsiflexor moment). The lower moment values for the knee and hip were attributed to a shorter stride length, a slower

All correspondence to be addressed to Edward Lemaire, Prosthetics and Orthotics Department, The Rehabilitation Centre, 505 Smyth Road, Ottawa, Ontario, Canada K1H 8M2.

walking speed, and an increase in double support and stance.

Another study which focused on the role of the non-amputated limb in trans-tibial amputee locomotion was performed by Hurley *et al.* (1990). The subjects from this study were all under 45 years of age and over half of the subjects were not long-term prosthesis users. Through examination of angle-angle diagrams for the ankle, knee, and hip joints, it was concluded that amputee gait patterns were more asymmetrical than non-amputees (since the joint angles for the sound leg were different from the joint angles for the prosthetic leg). Examination of horizontal joint reactions forces at the ankle and the knee showed that non-amputees had higher peak positive values than amputees. The lower amputee joint reaction forces were attributed to a lower push-off force from the amputated side and a slower walking speed. These results were consistent with the results from Lewallen *et al.* (1986).

Recently, trans-tibial amputee gait studies have focused on the contribution of energy storing feet to walking and running kinetics (Winter and Sienko, 1988; Czerniecki *et al.*, 1991; Torburn *et al.*, 1990; Barth *et al.*, 1992). Winter and Sienko examined gait patterns for 5 trans-tibial amputees who used a SACH (Single Axis Cushioned Heel) foot. Two subjects were re-tested with a uniaxial foot and one subject was tested with a Greissinger foot. All subjects demonstrated a greater than normal hip extensor moment from early stance to mid-stance to compensate for the below average push-off from the prosthetic leg. This hip extensor moment accounted for a quadricpes co-contraction over the same period, thereby compensating for the resulting knee flexor moment. All other moment and power patterns at the knee and hip were comparable with normal data. There was no difference between the ankle plantarflexor moment curves for the various prosthetic feet; however, the magnitude of these curves was approximately ⅔ of normal. The energy recovery for the uniaxial foot and Greissinger foot was found to be 20% and 30% respectively. The moment patterns at the knee and hip for the Greissinger fitting produced results which were closer to normal than those of the SACH and uniaxial foot.

Czerniecki *et al.* (1991) found results similar to Winter and Sienko during their investigation of running characteristics for five male trans-tibial amputees. The subjects ran along a 20m runway while using either a Flex-Foot, a SACH foot, or a Seattle foot. The SACH foot results were consistent with the findings of Winter and Sienko, although the amount of energy recovered was slightly higher (31%). The Flex-Foot trials produced a superior result in terms of approximating normal gait pattterns and energy recovery (84%) while the Seattle foot had a moderate effect on energy recovery (52%). The energy recovery capabilities of these prosthetic feet were found to be better than those of the traditional units; however, they were not comparable in respect of energy generation to normal plantarflexor activity (241% of the energy absorbed).

Torburn *et al.* (1990) used stride characteristics, joint kinematics, joint kinetics, electromyography (EMG), and physiological assessment to compare the Flex-Foot, STEN, Seattle, and Carbon Copy II (CCII) feet with the SACH foot. Kinematic test results showed that the CCII foot produced or permitted a significantly higher cadence and shorter gait cycle duration than the SACH and Flex-Foot ($p=0.02$). The relatively low statistical power, however, indicated that more subjects would be necessary to account for the majority of effects. Joint mechanical analysis demonstrated a larger ankle dorsiflexor moment at push-off from the Flex-Foot trials but no difference was found between any other prosthetic feet. The lack of a difference between the Seattle foot and SACH foot in terms of the dorsiflexor moment at push-off does not correspond with the results of Czerniecki *et al.* (1991). No differences were found for the EMG or physiological energy cost tests.

A study similar to Torburn *et al.* (1990) was performed by Barth *et al.* (1992). Gait kinematics, ground reaction forces, and physiological energy cost were used to assess the function of SACH, SAFE II, Seattle Lightfoot, Quantum, Carbon Copy II, and Flex-Walk feet. It should be noted that a treadmill was used for the energy cost protocol in this study. Since a treadmill can actively pull the support leg backward during gait, the gait pattern of these amputees could have been altered and the energy storing capabilities of these feet may not have been realised. These factors could account for the lack of difference

Table 1. Subject and prosthesis characteristics.

Subject	Age	Height (m)	Mass (kg)	Socket	Foot	Time Since Amputation (years)
1	69	1.80	88.6	PTB	SACH	47
2	67	1.79	80.5	PTB	SACH	46
3	71	1.73	86.4	PTB	SACH	46
4	68	1.71	80.0	PTS	SACH	43
5	66	1.81	82.3	PTS	Single axis	46
6	67	1.87	95.5	PTS	Multiaxis	46
7	70	1.75	77.3	PTS	Flex-Foot	46
8	72	1.72	78.9	PTS	Seattle	46

found between feet in the energy cost trials. The kinematic results showed a significant difference in linear velocity, cadence, stride length, and single limb stance time between the young, traumatic amputees and the older, vascular amputees ($p<0.1$). The Carbon Copy II and Quantum feet were shown to have significantly higher peak ground reaction force values at weight acceptance than in the other test cases ($p<0.1$). There was no significant difference between prosthetic feet for push-off forces.

Although these studies provided valuable information on trans-tibial amputee gait, the gait patterns of the elderly amputee have not received adequate attention. The majority of the present studies have also focused on the amputated side of the body. Due to the importance of the non-amputated limb for propulsion during locomotion (Seliktar and Mizrahi, 1986) examination of the kinematic and kinetic gait parameters for a group of elderly amputees is warranted.

Methods

Eight men with trans-tibial amputations who were over 65 years of age, had lost their leg due to trauma, and had worn a prosthesis for at least 25 years were recruited through the War Amputations of Canada. Before testing, all subjects were assessed by a prosthetist to ensure optimal fit and function of the prosthesis. The prosthetist also ensured that none of the subjects had stump problems (pain, swelling, pressure sores, etc.). The majority of the subjects had a socket with supracondylar suspension and used a SACH foot (Table 1). Anthropometric measurements were taken from the non-amputated side and body segment parameters were estimated using the relationships defined by Dempster (1955) and reported by Winter (1979).

The gait testing session involved placing joint markers on the shoulder, hip, knee, heel, ball of the foot, and toe of the subject's non-amputated side (Plagenhoef, 1971). Following a series of warm-up trials, the subjects walked at a natural cadence along a walkway while cinematographic and force plate data were collected at 50 Hz (Locam camera and Kistier force plate connected to a Data General mini computer). Six trials were recorded for each subject. The resulting data were corrected for perspective, filtered at 6 Hz, and used to calculate joint kinematics, net joint moments, and joint powers via the BIOMECH analysis package (Robertson and Winter, 1980; Lemaire and Robertson, 1989). The kinematic data were used to calculate stride length and walking speed.

The individual results were normalised to 100% of the stride time and to body mass before ensemble averages were calculated for each subject and for all subjects (grand ensemble). The ensemble averaged curves were

Table 2. Stride length, stride length/height ratio, and stride velocity averaged over individual trials.

Subject	Stride Length (m)	Stride Length / Height	Velocity (m/s)
1	1.21	0.67	0.95
2	1.41	0.79	0.97
3	1.37	0.79	1.27
4	1.34	0.75	1.13
5	1.57	0.86	1.46
6	1.48	0.79	1.16
7	1.47	0.84	1.33
8	1.43	0.84	1.32
ALL	**1.41**	**0.79**	**1.20**

used for all analyses and comparisons between the test results and normal data.

Results

Stride analysis

Table 2 displays the average stride length, stride length represented as a proportion of height, and walking velocity. The velocity values had a standard deviation of 0.18 and a range of 0.51 m/s.

The subjects who used energy storing feet and PTS suspension had higher walking velocities than all other subjects except for the subject with a single-axis foot and PTS suspension. It should be noted, though, that the faster walker was the youngest and the tallest subject in the study.

Joint moments and powers

Although a variety of prosthetic feet and suspension techniques were utilised by the subjects in this study, the gait patterns for the non-amputated limb were very similar. This similarity is reflected in the relatively low coefficient of variation (CV) values found for the grand ensemble average curves (Figs 1, 2 and 3).

A very consistent temporal relationship existed among trials and among subjects. The maximum range found between corresponding event codes (intersubject) was 0.08 s. This similarity permits inclusion of the ensemble averaged codes on Figures 1, 2, and 3.

The results from the comparison between existing data on non-amputees and the subjects from this study are listed in Tables 3, 4, and 5. The events referred to in the tables are marked on Figures 1, 2, and 3.

No substantial differences in the shape of the

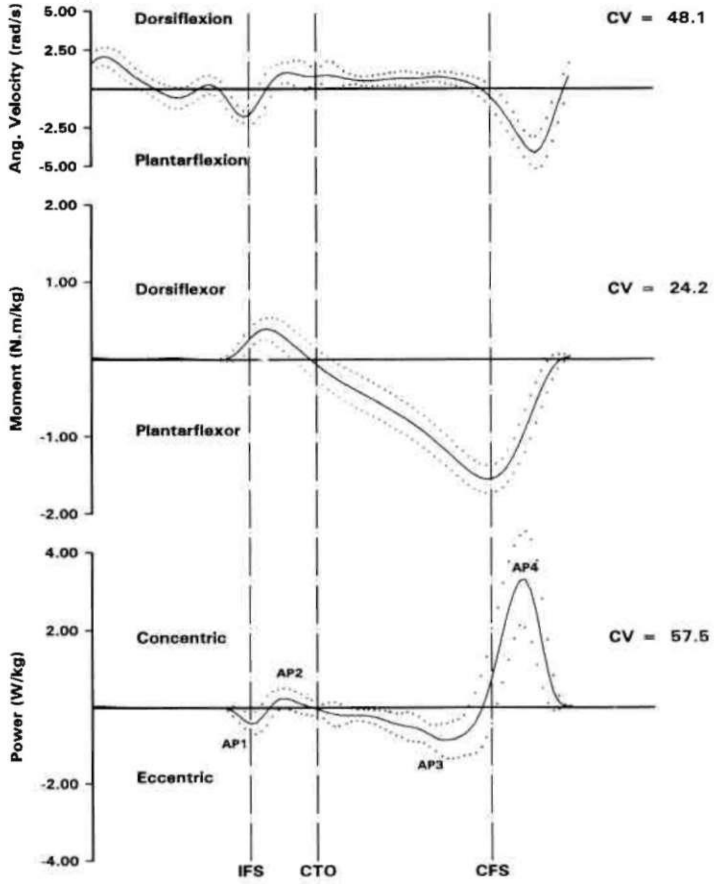

Fig. 1. Ensemble averages, standard deviations, and coefficients of variation for ankle angular velocity, moment, and power for all test subjects. The gait cycle started from non-amputated leg toe-off (IFS=ipsilateral foot-strike, CTO=contralateral toe-off, CFS=contralateral foot-strike).

moment and power curves were found during the between-subject evaluation. The magnitudes of the critical points along these curves were different; however, no relationship was found based on the type of suspension (PTB versus PTS) or the type of foot (SACH, multiaxis, energy storing). Generally, the slower walkers had lower moment and power values.

Discussion

Stride characteristics

It is generally assumed that the elderly walk more slowly and have a shorter stride length than the younger generation (Elble *et al.*, 1991; Murray *et al.*, 1969; Winter *et al.*, 1990), however, the amputees from this study had an average walking velocity and average stride length comparable to or above similar results from previous studies on younger amputees (Table 6). The results from Barth *et al.* (1992) involved three subjects who had an average age of 64.4 years and lost their leg as a result of vascular disease (average of five years since amputation). The extremely low results from this study may indicate a difference between elderly amputees who lost their leg due to trauma and people who had an amputation due to vascular disease. Another possible explanation is that a difference exists between long-term and short-term elderly prosthesis users.

The lack of a distinct age difference for walking speed or stride length may occur because the prime limitation for the amputee is the device and not physical capabilities which diminish with age. As technology provides more efficient means to store and/or generate energy on the prosthetic side, age related

Table 3. Gait characteristics at the ankle for all subjects (non-amputated limb). Comparisons are made with corresponding data for young non-amputees (†Winter, 1988) and elderly non-amputees (‡Winter, 1990).

Ankle				
	Event	Test Subjects	Young Non-amputees†	Elderly Non-amputees‡
Swing Phase	Concentric dorsiflexion (generation by dorsiflexors)	Small dorsiflexor moment to lift toe at initiation of swing phase (eliminate drop-foot).	Essentially no swing phase moment or power.	Essentially no swing phase moment or power.
	20% – 80% of swing phase	Essentially no ankle moment or power.	Essentially no swing phase moment or power.	Essentially no swing phase moment or power.
	Eccentric dorsiflexion (absorption by dorsiflexors)	Ankle resists plantarflexion as foot prepares for heel-strike.	Essentially no swing phase moment or power.	Essentially no swing phase moment or power.
Support Phase	Eccentric dorsiflexor (absorption by dorsiflexors)	Ankle resists plantarflexion to limit footslap at heel-strike.	Very small eccentric dorsiflexor moment.	Very small eccentric dorsiflexor moment.
	Concentric dorsiflexion (generation by dorsiflexors)	Ankle actively dorsiflexes until foot-flat to assist in moving the lower leg to mid-stance position and for stability.	Eccentric plantarflexor moment to control leg as it rotates over flat foot.	Very small concentric dorsiflexor moment.
	Eccentric plantarflexion (absorption by plantarflexors)	Ankle resists excessive dorsiflexion during mid-stance to control leg as it rotates over flat foot.	Same curves as normals but amputees have a larger peak power value.	Same curves as normals but amputees have a larger peak power value.
	Concentric plantarflexion (generation by plantarflexors)	Large plantarflexor moment and power during push-off.	Same curves as normals but amputees have a much larger peak power value.	Same curves as normals but amputees have a larger peak power value.
General				
– Amputee curves show much lower CV's than normals. This is contrary to expected results for amputees. – Essentially no ankle moment or power during the majority of swing phase. – Peak moments are similar to normals. – Peak powers larger in magnitude than normal during mid-stance and during push-off.				

differences in stride characteristics may become apparent. The tendency for the SACH foot users to have a slower walking speed than the other subjects may support this view, although the subjects who switched to energy storing prosthetic feet may have done so since they naturally walked faster or were in better physical condition. More research concerning energy storing prostheses and elderly amputee gait is required to determine the effectiveness of these devices on the senior population.

Gait

To determine the relevance of the results from amputee test trials, the test data were compared with the results from Winter (1988) and Winter *et al.* (1990). The earlier document by Winter provided gait results at three cadences, thereby facilitating comparison of relative values without accounting for walking speed. The 1990 document was used as a source for comparison since it contained moment and power results for elderly amputees (intra-subject data for a sample which averaged 68 years).

It is generally considered that the variability between amputee gait trials is greater than the variability between normal (Seliktar and Mizrahi, 1986). The CVs from this study contradicted this concept since the amputee trial variabilities were substantially lower than for the normal gait results. The low CV values may occur because the trans-tibial amputees were a more homogeneous group than the non-amputees and, since these elderly amputees were long-term prosthesis users, this group was

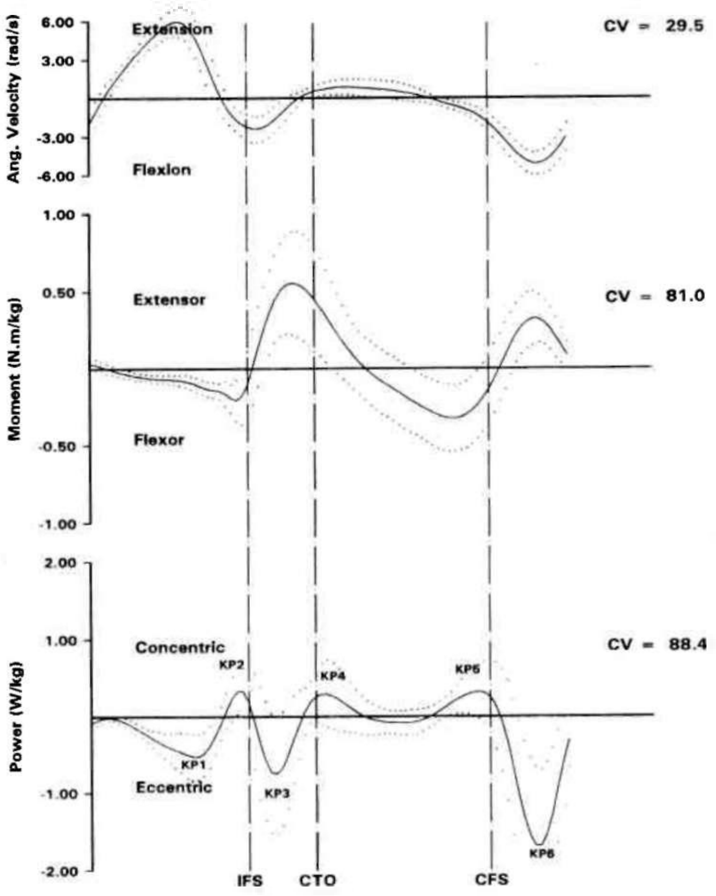

Fig. 2. Ensemble averages, standard deviations, and coefficients of variation for knee angular velocity, moment, and power for all test subjects. The gait cycle started from non-amputated leg toe-off (IFS=ipsilateral foot-strike, CTO=contralateral toe-off, CFS=contralateral foot-strike).

more homogeneous than the amputees who participated in previous studies.

Ankle

The shapes of the moment and power curves for the amputee subjects were essentially the same as those of results obtained from non-amputees (Winter, 1988), however, discrepancies were noted at the start of the stance phase. Normal subjects used the ankle dorsiflexors to absorb power at the beginning of stance while the amputee subjects had both power absorption (AP1) and power generation (AP2) over the same period. The concentric power burst may be required by the amputees to assist in "pulling" the lower leg segment through to mid-stance since the energy return capabilities of prosthetic legs will not produce the required moment at push-off. The lower push-off forces and moments from the prosthetic side have to be compensated by power generation from the dorsiflexors of the intact leg. The non-amputees may not require this concentric activity; therefore, they use the ankle dorsiflexors to control lower leg motion. The lack of two power bursts at the beginning of stance in the non-amputee data (Winter, 1988) may also be due to averaging of a group of non-homogeneous subjects, thereby masking small individual differences. In two studies using similar data collection and processing methods these two power events were present (Winter and Sienko, 1988; Winter *et al.*, 1990). These curves show an eccentric-concentric power pattern similar to the amputee subjects, although the magnitudes of these power bursts are much lower for the non-amputees. An eccentric dorsiflexor moment is initiated just before heel-strike for all amputee subjects but at heel-strike for elderly non-amputees. This may indicate a preparatory phase for the amputee group before weight is being transferred from the amputated side of the body to the non-amputated side.

The shapes of the moment and power curves during mid-stance and push-off were consistent

Table 4. Gait characteristics at the knee for all subjects (non-amputated limb). Comparisons are made with corresponding data for young non-amputees (†Winter, 1988) and elderly non-amputees (‡Winter, 1990).

Knee				
	Event	Description	Young Non-amputees†	Elderly Non-amputees‡
Swing Phase	Eccentric Flexor (absorption by flexors)	Knee flexors reduce the amount of knee extension during swing (prevent knee hyperextension).	Same curve shape, although the peak moment is higher for amputees.	Same curve shape, although the peak moment is higher for amputees.
	Concentric Flexor (generation by flexors)	Knee actively flexes to prepare for heel-strike (i.e., receiving load by *breaking the knee*).	Same as amputee subjects.	Same as amputee subjects.
Support Phase	Eccentric Extensor (absorption by extensors)	Knee controls flexion during mid-stance.	Same as amputee subjects.	Same as amputee subjects.
	Concentric Extensor (generation by extensors)	Knee actively extended to support bodyweight and raise centre of gravity at toe-off (amputated leg).	Same as amputee subjects.	Same as amputee subjects.
	Concentric Flexor (generation by flexors)	Knee actively flexes to prepare for push-off.	Same as amputee subjects.	Same as amputee subjects.
	Eccentric Extensor (absorption by extensors)	Knee extensors control knee flexion during push-off.	Similar curve but the amputee trials have a higher peak moment and power.	Similar curve but the amputee trials have a higher peak moment and power.

General

- Shape of moment and power curves comparable with Winter.
- Power curve has similar shape but is slightly offset temporally to the left (i.e. Winters KP3 curve occurs after the same curve for the amputees).
- Power and moment curves have higher magnitudes than normal during push-off.

for all available data but the moment and power values were higher for the amputee group. The larger power values for the AP3 power burst are used to control the amount of dorsiflexion before push-off. Controlled dorsiflexion is necessary to slow the forward progression of the tibia after mid-stance. This function is necessary to prepare for weight transfer to the prosthetic side.

The push-off power (AP4) was larger for the amputee group since the non-amputated leg must compensate for the lack of an energy generating segment on the amputated side. The amputee moment and power results were much larger than the elderly non-amputee results.

Knee

The moment and power curves for both groups were similar in shape but the peak values during swing and at push-off were higher for the amputees. The large eccentric flexor moment during swing may occur in response to a faster leg movement. Since the support time for the amputated side is less than the support time for the intact side the non-ampuated leg must complete the swing phase in less time in order to maintain a degree of symmetry. The resulting increase in speed would require a larger eccentric flexor moment to limit knee extension at the end of the swing phase.

The larger knee moment and power at push-off was in response to a larger concentric plantarflexor moment at the ankle and concentric flexor moment at the hip. Since peak powers at the hip and ankle were larger than for normals, a larger eccentric extensor moment was required to control the rate of knee flexion during push-off.

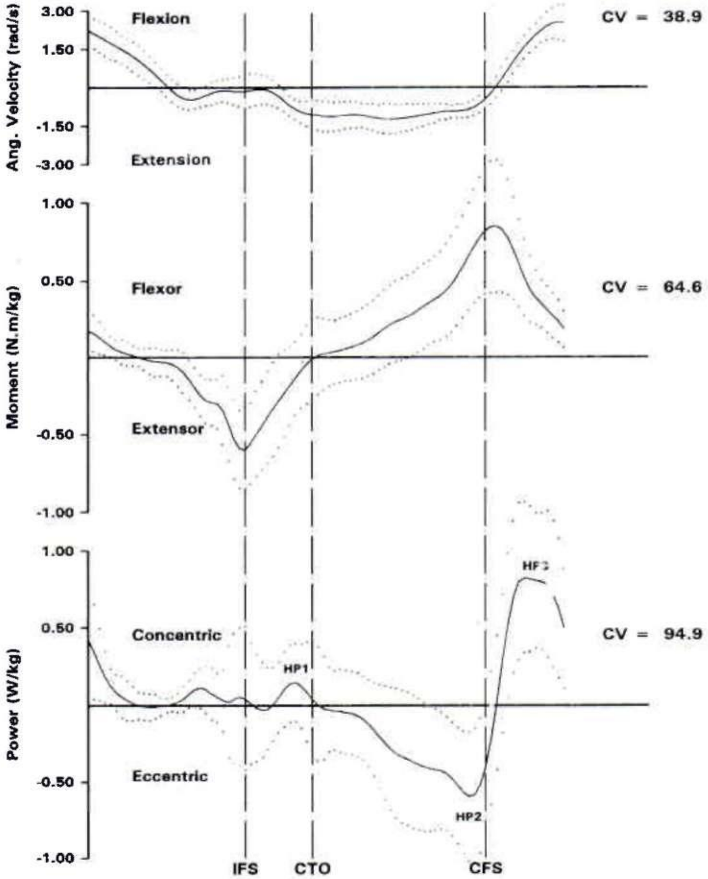

Fig. 3. Ensemble averages, standard deviations, and coefficients of variation for hip angular velocity, moment, and power for all test subjects. The gait cycle started from non-amputated leg toe-off (IFS=ipsilateral foot-strike, CTO=contralateral toe-off, CFS=contralateral foot-strike).

Hip

The shapes of the amputee hip moment and power curves were very similar to the reference curves for non-amputees; however, larger moment and power values were found during the support phase. A greater concentric flexor moment at push-off was necessary to initiate a faster swing phase (since the non-amputated leg has a shorter swing phase period than the amputated leg). The larger eccentric flexor moment just before push-off may be the result of the same mechanism as the eccentric extensor moment at the ankle (preparing for transfer of weight to the prosthetic limb). The hip flexors were required to control extension of the hip before active flexion was initiated at push-off.

Intersubject evaluation

The lack of intersubject difference between non-amputated leg gait patterns for PTB or PTS sockets was expected since a good fit with either design should provide a satisfactory interface for walking at a natural pace. The type of prosthetic foot did not appear to have an effect on the non-amputated leg; however, this study had an insufficient number of subjects to reach such a conclusion. Based on observations

Table 5. Gait characteristics at the hip for all subjects (non-amputated limb). Comparisons are made with corresponding data for young non-amputees (†Winter, 1988) and elderly non-amputees (‡Winter, 1990).

Hip				
	Event	Description	Young Non-amputees†	Elderly Non-amputees‡
Swing Phase	Concentric Flexor (generation by flexors)	Hip flexion at, and just after, toe-off to lift leg (allow foot to clear floor).	Same as amputee subjects.	Same as amputee subjects.
	Concentric Extensor (generation by extensors)	Hip actively extended during last half of swing.	Similar curve but the amputee trials have a slightly higher peak moment.	Same as amputee subjects.
Support Phase	Concentric Extensor (generation by extensors)	Hip extends at weight acceptance (pull leg into support position).	Same as amputee subjects.	Much longer concentric extensor moment than amputees (66% of stance phase).
	Eccentric Flexor (absorption by flexors)	Limit hip extension as thigh rotates backward after amputated leg push-off (prevent collapse at hip).	Similar curve from Winter but larger values for both moment and power.	Shorter duration than amputees.
	Concentric Flexor (generation by flexors)	Initiation of swing phase (moves leg upward and forward).	Similar curve but the amputees trials have a larger peak moment and power.	Same as amputee subjects.
General				
–	The moment and power curve for both the amputee groups and the normal groups have the same shape.			
–	Larger peak values were found for the amputees at the initiation of swing and during mid-support.			

Table 6. Comparison of average walking velocities and stride lengths.

Study	Subject	Subject Age (years)	Average Velocity (m/s)	Average Stride Length (m)
Doane and Holt (1983)	Amputee	55–67	1.22	
Gonalez *et al.* (1974)	Amputee	43–77	1.07	
Robinson *et al.* (1977)	Amputee	21–73	1.07	1.32
Torburn *et al.* (1990)	Amputee	39–57	1.17	1.40
Barth *et al.* (1992)	Amputee	36–67	0.75	1.10
Waters *et al.* (1988)	Non-amputee	60–80	1.19	1.27
This Study	**Amputee**	**66–72**	**1.20**	**1.41**

from the prosthetist, some elderly amputees may not deform the keel contained in energy storing feet to the same degree as younger amputees. The subject who used a Flex-Foot did not appear to make use of the energy return capabilities of this device during walking but used this foot because of its light weight. Additional research is required to determine if energy storing feet have an effect on the moments and powers for the non-amputated leg of elderly amputees.

Conclusion

Gait patterns from the non-amputated leg of experienced, elderly, men with trans-tibial amputations were shown to be comparable with data from non-amputees, although anomalies were found just after heel-strike and at push-off. All differences could be explained by the lack of an ankle moment generator on the amputated leg. The between-subject variability was much lower than expected, indicating a degree of homogeneity for long-term prosthesis users. Further research involving the elderly population is essential to quantify the benefits to them of modern prosthetic components and techniques.

Acknowledgements

The authors would like to acknowledge the clinical contribution of Sam Smith C.P. (c) and the administrative assistance of Guy Martel C.P.O. (c), M.H.Sc. towards completion of this project. The technical assistance of Mr. Wayne Albert is also acknowledged. This project was funded by the War Amputations of Canada.

REFERENCES

Barth DG, Schumacher L, Sienko-Thomas S (1992). Gait analysis and energy cost of below-knee amputees wearing six different prosthetic feet. *J Prosthet Orthot* **4**, 63–75.

Breakey J (1976). Gait of unilateral below-knee amputees. *Orthot Prosthet* **30(3)**, 17–24.

Czerniecki JM, Gitter A, Munro C (1991). Joint moment and muscle power output characteristics of below-knee amputees during running: the influence of energy storing prosthetic feet. *J Biomech* **24**, 63–75.

Dempster WT (1955). Space Requirements of the Seated Operator. Wright Patterson Air Force Base (WADC-TR.)

Doane NE, Holt LE (1983). A comparison of the SACH and single axis foot in the gait of unilateral below-knee amputees. *Prosthet Orthot Int* **7**, 33–36.

Elble RJ, Sienko-Thomas S, Higgins C, Colliver J (1991). Stride-dependent changes in gait of older people. *J Neurol* **238**, 1–5.

Enoka RM, Miller DI, Burgess EM (1982). Below-knee amputee running gait. *Am J Phys Med* **61**, 66–84.

Ganguli S, Mukherji P, Bose KS (1974). Gait evaluation of unilateral below-knee amputees fitted with patellar-tendon-bearing-prostheses. *J Indian Med Assoc* **63**, 256–259.

Gonzalez EG, Corcoran PJ, Reyes RL (1974). Energy expenditure in below-knee amputees: correlation with stump length. *Arch Phys Med Rehabil* **55**, 111–119.

Hannah RE, Morrison JB, Chapman AE (1984). Prosthetic alignment: effect on gait of persons with below-knee amputations. *Arch Phys Med Rehabil* **65**, 159–162.

Hunter G, Waddell J (1976). Management of the patient requiring leg amputation for peripheral vascular disease. *Can Med Assoc J* **115**, 634–638.

Hurley GRB, McKenney R, Robinson M, Zadravec M, Pierrynowski MR (1990). The role of the contralateral limb in below-knee amputee gait. *Prosthet Orthot Int* **14**, 33–42.

Lemaire ED, Robertson DGE (1989). Power in sprinting. *Track Field J* **35**, 13–17.

Lewallen R, Dyck G, Quanbury A, Ross K, Letts M (1986). Gait kinematics in below-knee child amputees: a force plate analysis. *J Pediatric Orthop* **6**, 291–298.

Murray MP, Kory RC, Clarkson BH (1969). Walking patterns in healthy old men. *J Gerontol* **24**, 169–178.

Plagenhoef S (1971). Patterns of Human Movement. – Englewood Cliffs: Prentice Hall. p11-14.

Robinson JL, Smidt GL, Arora JS (1977). Accelerographic, temporal, and distance gait factors in below-knee amputees. *Phys Ther* **57**, 898–904.

Robertson DGE, Winter DA (1980). Mechanical energy generation, absorption and transfer amongst segments during walking. *J Biomech* **13**, 845–854.

Seliktar R, Mizrahi J (1986). Some gait characteristics of below-knee amputees and their reflection on ground reaction forces. *Eng Med* **15**, 27–34.

Torburn L, Perry J, Ayyappa E, Shenfield SL (1990). Below-knee amputee gait with dynamic elastic response prosthetic feet: a pilot study. *J Rehabil Res Dev* **27**, 369–384.

WATERS RL, LUNSFORD B, PERRY J, BYRD R (1988). Energy-speed relationship of walking: standard tables. *J Orthop Res* **6**, 215–222.

WINTER DA (1979). Biomechanics and human movement. – Toronto: John Wiley and Sons. p150–152.

WINTER DA (1988). The biomechanics and motor control of human gait. – Waterloo; University of Waterloo Press.

WINTER DA, SIENKO SE (1988). Biomechanics of below-knee amputee gait. *J Biomech* **21**, 361–367.

WINTER DA, PATLA AE, FRANK JS, WALT SE (1990). Biomechanical walking pattern changes in the fit and healthy elderly. *Phys Ther* **70**, 340–347.