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The role of the contralateral limb in below-knee amputee gait

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Abstract

Very little quantitative biomechanical research has been carried out evaluating issues relevant to prosthetic management. The literature available suggests that amputees may demonstrate an asymmetrical gait pattern. Furthermore, studies suggest that the forces occurring during amputee gait may be unequally distributed between the contralateral and prosthetic lower limbs. This study investigates the role of the contralateral limb in amputee gait by determining lower limb joint reaction forces and symmetry of motion in an amputee and non-amputee population. Seven adult below-knee amputees and four non-amputees participated in the study. Testing involved collection of kinematic coordinate data employing a WATSMART video system and ground reaction force data using a Kistler force plate. The degree of lower limb symmetry was determined using bilateral angle-angle diagrams and a chain encoding technique. Ankle, knee and hip joint reaction forces were estimated in order to evaluate the forces acting across the joints of the amputee's contralateral limb. The amputees demonstrated a lesser degree of lower limb symmetry than the non-amputees. This asymmetrical movement was attributed to the inherent variability of the actions of the prosthetic lower limb. The forces acting across the joints of the contralateral limb were not significantly higher than that of the non-amputee. This suggests that, providing the adult amputee has a good prosthetic fit, there will not be increased forces across the joints of the contralateral limb and consequently no predisposition for the long-term wearer to develop premature degenerative arthritis.

Introduction

There appears to be an increase in the number of below-knee amputees in our population due to ageing, accidents, and surgery related to peripheral vascular disease. This increase in the amputee population warrants research that attempts to address issues relevant to prosthetic management. It is clinically valuable to understand the role of the contralateral limb in amputee gait since, if the joint forces in the contralateral limb exceed natural limits, the individual may be predisposed to premature degenerative arthritis (Lewallen et al., 1986). In an attempt to equalize step length, improve balance and ensure knee stability, the prosthetist strives to achieve a symmetrical gait pattern when aligning and fitting an amputee with a prosthesis. Evaluating lower limb symmetry may therefore contribute to a better understanding of the role of the contralateral limb. The purpose of this study is to investigate the role of the contralateral limb in amputee gait by determining lower limb joint reaction forces and symmetry of motion in an amputee and non-amputee population.

Amputee gait has been evaluated both qualitatively (Gonzalez et al., 1974; Urban 1973; Waters et al., 1976; Kegel et al., 1981) and quantitatively in the literature. Quantitative research can be further subdivided into kinematic and kinetic studies. Many studies have presented descriptions of amputee gait based on kinematic measures (Enoka et al., 1982; Hannah and Morrison, 1984; Zuniga et al., 1972). James and Oberg (1973) and Murray et al. (1981), in similar studies of above-knee amputees, found

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significant differences in velocity, cadence, gait cycle, and stride length when their study group was compared to normal subjects. These authors noted that the step length of the prosthetic leg tended to be longer than that of the contralateral leg. Breakey (1976), in studies of below-knee amputees, reported that the stance phase of gait was longer in the normal limb and shorter in the amputated limb. Robinson et al. (1977) obtained time distance and accelerometer data from below-knee amputees. The subjects took longer steps more quickly on their prosthetic side and the resulting gait was described as asymmetric. Hershler and Milner (1980) also found asymmetry between the unaffected and the amputated side when looking at the variation of hip angle and knee angle throughout all phases of gait in above-knee amputees. Skinner and Effeney (1985) noted that this asymmetry of motion increases the excursion of the centre of mass during each cycle and thereby increases the energy cost of ambulation. These kinematic studies suggest that amputees demonstrate an asymmetric gait pattern. However this observation has not been verified using quantitative methods to determine the degree of symmetry.

Recently, Middleton et al. (1988) used lower limb symmetry as a criterion for evaluating the effects of a rigid ankle-foot orthosis and a hinged ankle-foot orthosis on a spastic diplegic cerebral palsied child. Kinematic and kinetic variables were determined using a video acquisition system and a Kistler force plate. Employing lower limb angle-angle diagrams and a chain encoding technique (McIlwain and Jensen, 1985), differences in lower limb symmetry while unbraced, and in the braced conditions were determined.

Very little quantitative biomechanical literature is available that evaluates the mechanics of amputee gait utilizing kinetic analyses (Cappozzo et al., 1976; Golbranson, 1980; Lewallen et al., 1985). The majority of this research focuses on evaluating different prosthetic components with regard to amputee gait (Clark and Zernicke, 1981; Hoy et al., 1982; Zernicke et al., 1985). Winter and Sienko (1988) used sagittal plane biomechanical and EMG analyses from eight below-knee amputee trials to demonstrate modified motor patterns from the residual muscles at the hip and knee. Seven of the eight amputee trials were with SACH feet and showed a negligible knee moment of force during early stance (when non-amputees show an extensor moment), and a below normal knee moment of force in late stance. They explain that because of hyperactivity of the hamstrings during early stance there is an excessive knee flexor moment which is cancelled out by co-contracting knee extensors at that time.

Suzuki (1972) used a force plate to measure the three dimensional ground reaction forces on the limb during stance phase. He found the vertical ground reaction forces for the prosthetic and contralateral limbs to be different in subjects with below-knee, above-knee and hip disarticulation amputations. The vertical ground reaction force measured in below-knee amputees for the prosthetic limb was lower in magnitude with a smaller trough than the ground reaction forces measured on the contralateral side. Oberg and Lanshammar (1982) used a SELSPOT motion analysis system and force plate to study amputee gait. Knee moments and gait pattern-force vector diagrams were reported for one above-knee amputee. The authors noted differences between the subject's prosthetic and contralateral sides, however, they were only able to conclude that this type of analysis is valuable in evaluating amputee gait.

Lewallen et al. (1986) have produced the only study evaluating the development of amputee gait in children with respect to potential long term influences. This study compared kinematic and kinetic parameters of a normal and amputee paediatric population (6 amputees, 6 non-amputees) in an attempt to determine whether the loss of a limb segment results in increased forces across the intact joints of the normal limb. Quantitative analysis involved integration of force plate and cine data, and the inverse dynamics approach was utilized to estimate the joint moments in the intact limb. The authors reported that the normal leg in the child amputee displays reduced action and forces in order to achieve a better symmetry with the amputated leg. Furthermore, a tendency for the intact limb to have reduced forces involved in initial weight acceptance on the amputated limb was noted. It was concluded that the intact limb does not develop increased forces in the joints as compared with values for normals. This balance in the child amputee was achieved through slower walking velocity, decreased step length, and increased double support and stance phases as compared

TABLE 1 — Subject Demographics (A — amputee, S — non-amputee).

Subject	Age (yrs)	Height (M)	Weight (Kg)	Socket Design	Foot Component	Gait(2)	Amputation Year	Amputation Reason
A1	42	1.75	84.0	PTB(1)	Seattle(L)	Good	1973	Traumatic
A2	32	1.83	86.5	PTB	Flex(L)	Good	1984	Traumatic
A3	32	1.68	70.0	PTB	Seattle(R)	Good	1986	Congenital
A4	32	1.68	64.5	PTB	Seattle(R)	Fair	1987	Traumatic
A5	43	1.67	73.0	PTB	Seattle(R)	Excellent	1957	Traumatic
A6	42	1.81	98.0	PTB	Flex(R)	Fair	1986	Vascular
A7	26	1.70	60.0	PTB(1)	Seattle(L)	Good	1966	Traumatic

Subject	Age (years)	Height (M)	Weight (kg)
S1	26	1.77	71.0
S2	24	1.78	77.4
S3	27	1.80	81.1
S4	24	1.63	62.7

PTB = patellar-tendon bearing
(R) = right
(L) = left
(1) thigh corset & external hinges
(2) clinical subjective gait analysis

to his normal counterpart. The researchers concluded that, providing the child amputee has a good prosthetic fit, there will be no increased forces across the joints of the intact limb and consequently no predisposition towards premature degenerative arthritis. The conclusion drawn from this investigation is suspect since no statistical technique was employed when comparing joint forces between the amputee and non-amputee groups. These results are also limited since only one stride per subject was analyzed. Considering the supposed asymmetrical nature of amputee gait multi-stride analyses are warranted.

Methodology

Seven below-knee amputees and four non-amputees participated in the study. Information describing the subjects is presented in Table 1. None of the seven amputees was undergoing clinical prosthetic management at the time of testing. Prior to testing, prosthetic fit was checked and the amputee was questioned regarding his/her evaluation of prosthetic fit. All of the amputees reported that they were satisfied with their present prosthesis. During testing, the amputee's gait was clinically evaluated and characterized as either poor, fair, good, or excellent. All amputee subjects were younger than 45 years of age since the ramifications of long-term wear were of interest. None of the subjects had other medical conditions which could potentially affect their performance during testing. Four non-amputee subjects were selected to obtain data representative of non-amputee gait. The subjects were voluntary participants and informed consent was obtained prior to testing.

Data collection

Testing involved collection of kinematic coordinate data using a WATSMART video system and ground reaction force data employing a Kistler force plate. Anthropometric measurements of each subject were taken in order that segment inertial parameters could be estimated. Force plate and kinematic coordinate data was collected for the left lower limb of the non-amputee subjects and the contralateral limb of the amputee subjects. Kinematic coordinate data was collected for the right lower limb of the non-amputee subjects and the prosthetic lower limb of the amputee subjects. An independent three segment link system was used to model the motion of the contralateral/left lower limb during ambulation. Since only kinematic data was being collected on the prosthetic/right lower limb an independent two segment link system was used to model the motion of this side.

A four camera WATSMART kinematic data acquisition system was used to acquire the two-dimensional positions of five anatomical landmarks on the left/contralateral lower limb and three anatomical landmarks on the right/prosthetic side (50 hertz sampling rate). Eight one centimetre diameter disks containing 3 infra-red emitting diodes (IREDs) were placed over the anatomical landmarks. These landmarks approximated the positions of the anatomical joint centres of the hip, knee and ankle, and the proximal and distal ends of the foot segment on the left/contralateral lower extremity. IREDs were placed over the anatomical joint centres of the hip, knee and ankle of the non-amputee's right lower limb. The amputee's prosthetic lower limb was treated in a similar manner with the anatomical joint centres of the hip and knee located and a third IRED placed distally bisecting the longitudinal axis of the prosthetic shank in the sagittal plane. The cameras were placed perpendicular to the sagittal motion of the lower limbs. The subject was familiarized with the testing area in order to promote natural performances during data collection. Walking trials lasted 6-7 seconds and approximately 20 trials per subject were carried out.

Estimation of segment inertial parameters (mass and moment of inertia about the transverse proximal axis) were determined mathematically using regression equations (Jensen and Wilson, 1988). These regression equations employ selected anthropometric measurements (Hanavan, 1964) as predictor variables. Segment inertial parameters for the thigh, leg and foot of the amputee's contralateral limb and non-amputee's left limb were calculated in this manner.

Subsequent kinematic and kinetic analyses of coordinate data records were performed using the Waterloo Biomechanical Motion Analysis Software Package. The first eight trials in which the subject successfully contacted the force plate were used for analysis. Each walking trial was composed of one complete stride and both left and right lower limbs were analyzed. All joint coordinate data were filtered through a low pass recursive second order Butterworth digital filter using 5 hertz cutoff frequencies (Pezzack, 1977). Waterloo Programme input parameters were selected and employed in the established manner (Winter, 1979).

Statistical analysis

A chain encoding technique was used to quantify the degree of lower limb symmetry displayed by the subjects during ambulation (McIlwain and Jensen, 1985). This technique may be used to determine the degree of congruity between any two XY patterns. Each XY data set is converted into a chain encoded data set. The chain encoded data consists of a numeric array of single digits (0-7) describing the direction followed by straight lines connecting the original XY data points plotted on a square aspect ratio XY graph. Cosine cross-correlation analysis is used to determine the degree of congruity between the two chain encoded data sets. The cross-relation function derived from these two generated chains, referred to as the recognition

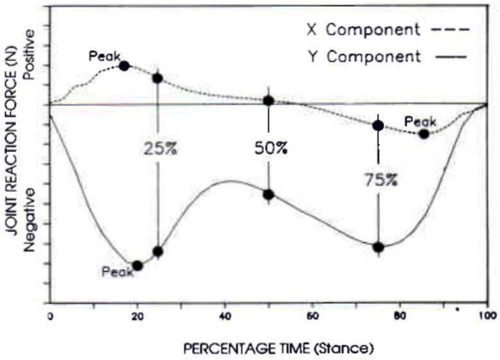

Fig. 1. Joint reaction force dependent variables.

coefficient, was used to quantify the degree of lower limb symmetry.

A 2 × 8 (subject × walking trial) partially repeated two-way analysis of variance (ANOVA) design was used to determine if the joint reaction forces acting on the contralateral limb of the amputee were significantly greater than those on the non-amputee during ambulation. Nine values of ankle, knee and hip joint reaction force, during the support phase of walking, were selected as dependent variables. All joint reaction force variables were normalized with respect to the subject's body mass. The nine dependent variables (depicted in Figure 1) per joint are as follows:

1) maximum positive horizontal joint reaction force,
2) maximum negative horizontal joint reaction force,
3) maximum vertical joint reaction force,
4) horizontal joint reaction force at 25% of stance,
5) horizontal joint reaction force at 50% of stance,
6) horizontal joint reaction force at 75% of stance,
7) vertical joint reaction force at 25% of stance,
8) vertical joint reaction force at 50% of stance,

and 9) vertical joint reaction force at 75% of stance.

This design strategy was employed so that peak values as well as the forces occurring during the natural progression through stance could be evaluated. In all 27 ANOVAs, walking trial was the within factor and subject was the grouping factor. The within factor was employed in order to

TABLE 2 — Chain encoding results of angle-angle plots for non-amputee walking trials.

Subject	S2 Right	S3 Right	S4 Right	S1 Left	S2 Left	S3 Left	S4 Left
S1 Right	0.851	0.847	0.767	0.871	0.831	0.836	0.762
S2 Right		0.868	0.818	0.837	0.878	0.875	0.835
S3 Right			0.792	0.840	0.869	0.897	0.776
S4 Right				0.739	0.850	0.787	0.879
S1 Left					0.831	0.828	0.748
S2 Left						0.860	0.827
S3 Left							0.800

determine if any testing effects (ie fatigue, practice) were present. Subjects were divided into two groups — amputee and non-amputee. Statistical analysis was used to compare the joint reaction forces occurring in the amputee's contralateral limb and the non-amputee's during ambulation.

Discussion

This design strategy was employed so that peak values as well as the forces occurring during the natural progression through stance could be evaluated. In each of the 27 ANOVAs, walking trial was the within groups factor and subject was the between groups factor (Winer, 1971). The within factor (walking trial) was employed in order to determine if any testing effects (ie fatigue, practice) were present. Subjects were divided into two groups — amputee and non-amputee. Statistical analysis was used to compare the joint reaction forces occurring in the amputee's contralateral limb during ambulation to the non-amputee's.

Normal gait is characterized by symmetrical movements of the lower limbs throughout the gait cycle. By adopting such a gait pattern an energy efficient mode of ambulation is obtained (Skinner and Effeney, 1985). Furthermore the forces during weightbearing are distributed equally between both lower limbs. As the degree of lower limb symmetry decreases, it is possible that the forces during weightbearing may become unbalanced between the hip, knee, and ankle joints of both lower limbs. This study attempts to quantify the degree of lower limb symmetry since it has been proposed that amputees demonstrate an asymmetrical gait pattern.

Lower limb symmetry

Angle-angle diagrams traditionally depict two selected lower limb joint angle variations plotted against each other for corresponding instants of time (McIlwain and Jensen, 1985; Hershler and Milner, 1980). For the purposes of this study, it was felt that absolute angular displacements of the thigh and leg segments best depicted the action of the lower limbs during ambulation. The absolute angular displacement of a limb segment is the inclination of this segment relative to the ground. Of the eight available strides, each subject's fourth walking trial was evaluated with regard to lower limb symmetry. Bilateral leg/thigh angle-angle diagrams were utilized in evaluating the degree of symmetry between the lower limbs. An estimate of congruity or similarity in shape between any two angle-angle configurations was obtained by chain encoding each pattern and then determining the cross-relation function from the two generated chains. This recognition coefficient (C) served as the criterion for intercurve comparisons (degree of symmetry). The recognition coefficient can vary from 0.0 to 1.0,

TABLE 3 — Chain encoding results of angle-angle plots for amputee walking trials: Contralateral versus prosthetic side (c-contralateral, p-prosthetic).

Subject	A1–P	A2–P	A3–P	A4–P	A5–P	A6–P	A7–P
A1–C	0.806	0.768	0.825	0.711	0.788	0.763	0.791
A2–C	0.855	0.798	0.844	0.789	0.849	0.816	0.797
A3–C	0.805	0.695	0.856	0.796	0.718	0.681	0.797
A4–C	0.836	0.801	0.828	0.735	0.828	0.803	0.806
A5–C	0.863	0.846	0.813	0.753	0.858	0.837	0.821
A6–C	0.861	0.796	0.852	0.773	0.821	0.768	0.850
A7–C	0.799	0.714	0.855	0.758	0.752	0.712	0.792

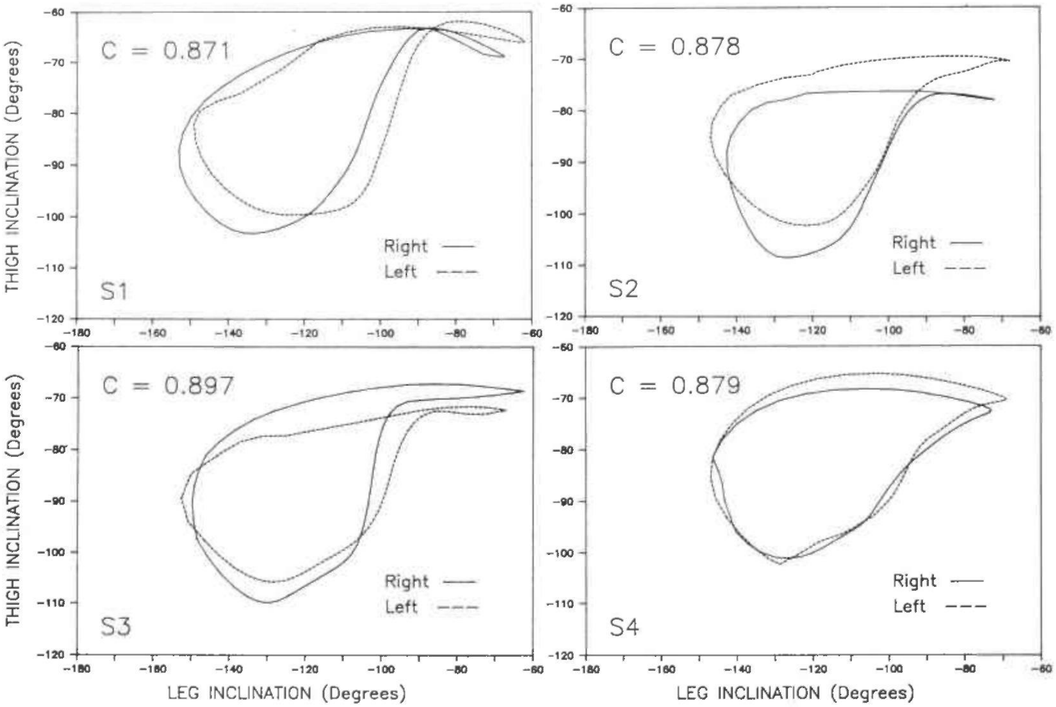

Fig. 2. Bilateral angle-angle plots of leg and thigh for one stride by non-amputees S1, S2, S3 and S4.

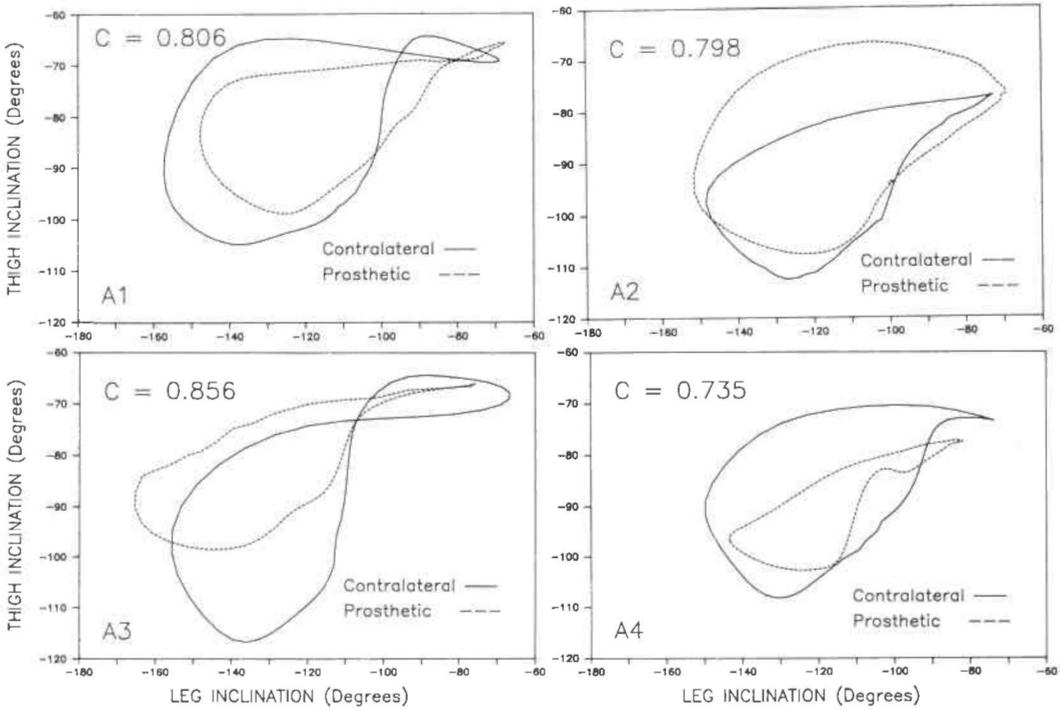

Fig. 3. Bilateral angle-angle plots of leg and thigh for one stride by amputees A1, A2, A3 and A4.

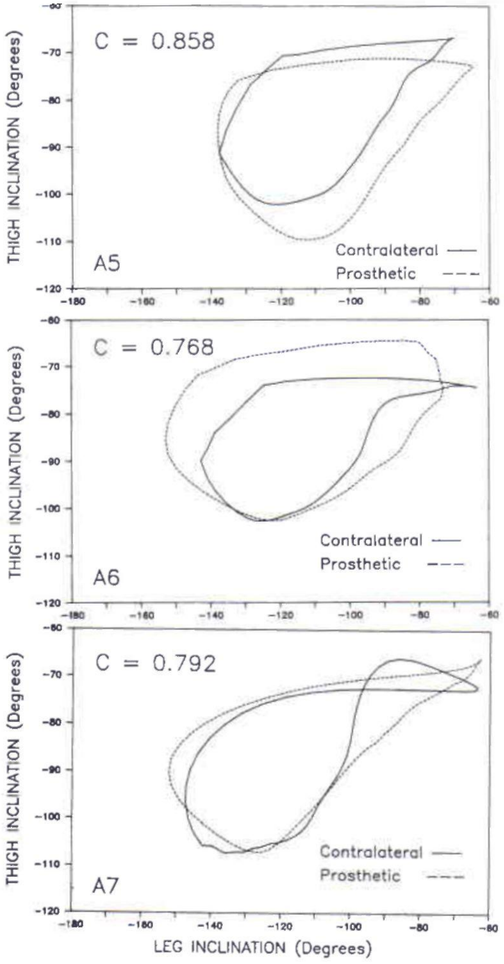

Fig. 4. Bilateral angle-angle plots of leg and thigh for one stride by amputees A5, A6 and A7.

with a value of 1.0 indicating perfect congruity and a value of 0.0 indicating absence of congruity between patterns, (McIlwain and Jensen, 1985). This chain encoding technique was employed in determining the degree of lower limb symmetry for both the amputee and non-amputee groups.

Tables 2 and 3 present the results of chain encoding both right/left and prosthetic/contralateral angle-angle plots for amputee and non-amputee subjects respectively (emboldened numbers). The non-emboldened values indicate between — subject variability for both groups. Figures 2 to 4 illustrate bilateral angle-angle diagrams where S1 through S4 represent the non-amputee group and A1 through A7 represent the amputee group. Non-amputees exhibited the highest degree of lower limb symmetry (mean 0.881, s.d. 0.011), whereas amputees demonstrated a lower degree of lower limb symmetry (mean 0.802, s.d. 0.044). Recognition coefficients, indicating degree of lower limb symmetry, ranged from 0.871 to 0.897 for non-amputees and from 0.735 to 0.858 for amputees. These results indicate that no amputee displayed a degree of lower limb symmetry equal to that of any non-amputee. The amputee demonstrating the highest recognition coefficient (A5, C = 0.858) has been a long time prosthetic wearer and was observed during testing to be an excellent walker. Amputee A4 displayed the lowest degree of lower limb symmetry (C = 0.735). Evaluation of bilateral angle-angle plots (Figure 3, subject A4) indicates limited movement on the prosthetic side suggesting a stiff-legged gait. It is interesting to note that this subject became an amputee quite recently (Table 1).

After evaluating all sound limb angle-angle diagrams for both the amputee and non-amputee groups it appears there exists a resemblance in the shape of the patterns between subjects. Conversely, the angle-angle diagrams depicting prosthetic side motion demonstrate a much more varied pattern between subjects. Tables 4 and 5 present recognition coefficient values evaluating between subject variability for contralateral and prosthetic sides, respectively. Between subjects, the contralateral limb exhibits a higher degree of lower limb symmetry (mean 0.833, s.d. 0.032) than the prosthetic side (mean 0.799, s.d. 0.054). The movements of the prosthetic lower limb may be characterized as more variable between subjects. These results indicate that non-amputees walk more symmetrically than amputees since movements of the prosthetic side do not mirror the sound or contralateral counterpart as well as a sound limb would.

Joint kinetics

Figure 1 illustrates a typical pattern of the horizontal and vertical components of the resultant force acting on any lower limb joint during normal level walking (Winter, 1987). During walking, the peak positive horizontal component of joint reaction force on the lower limb corresponds to weight acceptance and is initially forward in direction until approximately midstance. The negative horizontal component of joint reaction force occurs from approximately

TABLE 4 — Chain encoding results of angle-angle plots for amputee walking trials: Contralateral sides only.

Subject	A2	A3	A4	A5	A6	A7
A1	0.813	0.789	0.853	0.834	0.852	0.870
A2		0.809	0.844	0.868	0.847	0.816
A3			0.776	0.771	0.811	0.875
A4				0.848	0.842	0.840
A5					0.869	0.863
A6						0.820

Mean = 0.833 S.D. = 0.032

midstance to toe off with the peak corresponding to maximal forces acting during push off. The vertical component of the joint reaction force remains negative in direction (downward) throughout stance.

The results of the twenty-seven 2 × 8 (subject × walking trial) partially repeated two-way analyses of variance may be categorized into two areas: subject main effects and trial main effects. No significant trial main effect was displayed in any of the ANOVAs. This indicates that the likelihood of any error attributed to a testing effect, such as fatigue or practice, is negligible. No interactions between subject and trial were found.

Four significant subject (amputee/non-amputee) main effects were displayed in relation to the knee and ankle joint ANOVAs. No significant subject main effects were present for any ANOVA involving a hip joint reaction force dependent variable. The non-amputee demonstrated significantly higher peak positive horizontal components of joint reaction forces than the amputees in the study at both the ankle ($F(1,9) = 8.19$, $p < 0.05$) and knee ($F(1,9) = 10.26$, $p < 0.05$) joints. These effects were also demonstrated with regard to the values of ankle ($F(1,9) = 10.29$, $p < 0.05$) and knee ($F(1,9) = 7.13$, $p < 0.05$) horizontal component of joint reaction force occurring at 25% of stance. Figure 5 displays the ankle and knee horizontal components of joint reaction force occurring during stance for an amputee and non-amputee subject. Considering that peak horizontal joint reacton force occurs very close to the value corresponding to 25% of stance, it is understandable that significant effects were present for both of these dependent measures. The amputees experienced significantly lower ankle and knee horizontal components of joint

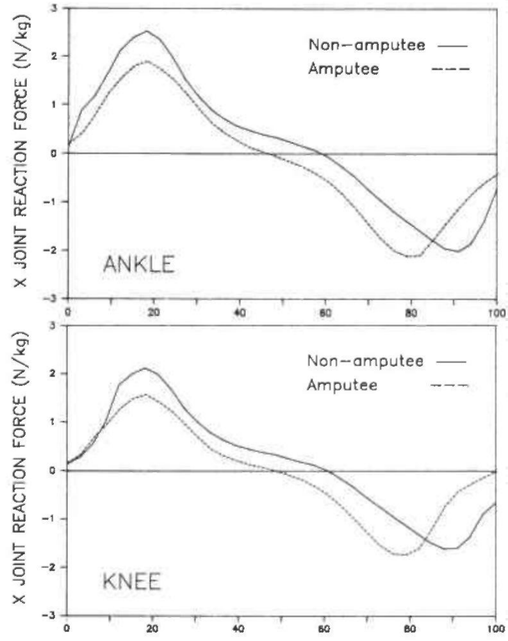

Fig. 5. Top, ankle horizontal joint reaction force, during stance, for an amputee and non-amputee subject. Bottom, knee horizontal joint reaction force, during stance, for an amputee and non-amputee subject.

TABLE 5 — Chain encoding results of angle-angle plots for amputee walking trials: Prosthetic sides only.

Subject	A2	A3	A4	A5	A6	A7
A1	0.798	0.812	0.765	0.846	0.802	0.875
A2		0.738	0.721	0.853	0.861	0.832
A3			0.806	0.766	0.745	0.834
A4				0.711	0.709	0.767
A5					0.857	0.874
A6						0.809

Mean = 0.799 S.D. = 0.054

TABLE 6 — Walking Velocity (ms^{-1}).

Subject	Trial 1	Trial 2	Trial 3	Trial 4	Trial 5	Trial 6	Trial 7	Trial 8	Mean
S1	1.505	1.372	1.498	1.474	1.435	1.396	1.417	1.428	1.44
S2	1.367	1.434	1.400	1.378	1.413	1.415	1.451	1.437	1.41
S3	1.364	1.434	1.524	1.451	1.537	1.535	1.529	1.446	1.48
S4	1.713	1.595	1.935	1.486	1.937	1.784	1.850	1.477	1.72
Non-amputees									1.51
A1	1.223	1.335	1.445	1.196	1.652	1.237	1.260	1.332	1.34
A2	1.442	1.524	1.525	1.551	1.595	1.648	1.510	1.448	1.53
A3	1.408	1.547	1.760	1.639	1.441	1.523	1.495	1.443	1.53
A4	1.118	1.046	1.154	1.219	1.275	1.208	1.171	1.151	1.17
A5	1.293	1.251	1.288	1.270	1.232	1.268	1.143	1.238	1.25
A6	1.169	1.099	1.071	1.147	1.160	1.041	1.023	1.169	1.11
A7	1.338	1.306	1.410	1.351	1.282	1.215	1.243	1.253	1.30
Amputees									1.32

reaction force on their contralateral side, at weight acceptance, than non-amputees. This may be due to the less active push-off inherent to prosthetic componentry on the amputee's prosthetic side as compared to his non-amputee counterpart. Decreased mass on the prosthetic side, relative to an intact lower extremity, might also contribute to the lower horizontal component of joint reaction force displayed by the amputees during weight acceptance.

The results indicate that the amputees evaluated in this study did not experience increased forces across the joints of their contralateral limbs as compared to a group of non-amputees. These results are in agreement with the findings of Lewallen and colleagues (1986). These researchers also reported that the child amputee accomplished this balance by walking slower than his non-amputee counterpart. Table 6 presents the walking velocity for all subject trials analyzed. Differences in walking velocity between the amputee (mean = 1.32 ms^{-1}) and non-amputee (mean = 1.5 ms^{-1}) groups were present. Furthermore, 5 of the 7 amputees average walking velocity over 8 trials was lower than the slowest non-amputee (8 trial average). It appears that the adult amputee may employ a slower walking velocity, than his non-amputee counterpart, in order to decrease the forces acting across the joints of his contralateral limb.

Conclusions

It has been proposed that amputees demonstrate an asymmetrical gait pattern with regard to lower limb movement. This statement is supported in this study since the amputees demonstrated a lesser degree of lower extremity symmetry than the non-amputees. It is proposed that this asymmetrical movement may be attributed to the inherent variability of the actions of the prosthetic lower limb. Although amputees may demonstrate an asymmetrical gait pattern, it appears that the forces acting across the joints of the contralateral limb are not significantly higher than that of a non-amputee. This being the case, providing the adult amputee has a good prosthetic fit, there will be no increased forces across the joints of the contralateral limb and consequently no predisposition for the long-term wearer to develop premature degenerative arthritis.

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