Scand J Rehab Med 29: 75–79, 1997

# DOUBLE-LIMB SUPPORT AND STEP-LENGTH ASYMMETRY IN BELOW-KNEE AMPUTEES

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**ABSTRACT. The sequence of gait events and symmetry of kinematic parameters between both lower limbs are compromised in below-knee amputees. In the present study the periods of double-limb support and the step length in below-knee amputees were investigated. The symmetry of the two periods of double-limb support occurring in each stride was obviously abnormal (ratio: 0.74) among temporal and distance parameters. The time of double-limb support ($0.211 \pm 0.05$ seconds) measured from heel-strike of the amputated leg until toe-off of the normal leg was significantly longer ($p = 0.011$) when compared with the contralateral leg ($0.173 \pm 0.04$ seconds). The step length of the normal leg ($0.709 \pm 0.07$ m) was significantly ($p = 0.045$) shorter than that of the amputated leg ($0.752 \pm 0.08$ m). Most of these differences between measured kinematic parameters can be explained by the limited ability of the prosthesis ankle-foot component to reproduce the normal functions of both foot and ankle.**

*Key words:* below-knee amputees, double-limb support, gait analysis, step length.

## INTRODUCTION

It is generally accepted that normal gait includes three basic tasks: weight acceptance, single limb support, and limb advance. It is also possible to distinguish eight separate phases in these tasks: initial contact, loading response, midstance, terminal stance, pre-swing, initial swing, midswing, and terminal swing (4, 6, 13). There are also two periods of double-limb support (DLS) in each gait cycle, which occur when both limbs support body weight simultaneously. Heel-strike initiates the period of DLS while toe-off of the controlateral leg terminates it. Symmetrical gait movements allow an energy efficient mode of ambulation (23) and were found to be a reliable indicator of gait training progression (2, 8, 17). Measurements of symmetry of temporal-distance variables during free-speed ambulation indicated that the highest value of symmetry observed was for step length (ratio: 0.98), followed by the stance time (ratio: 0.96) and the double-limb support time (ratio: 0.90; ratio: 0.99 for slow speed) (7).

Asynchronized sequence of gait phases and asymmetry of kinematic variables have been extensively investigated in below-knee (BK) amputees. Stance phase was found to be longer in the normal leg (3); and step length was longer in the amputated leg, but accomplished in less time than in the opposite side (15). Others, however, have reported that step length was longer in the normal leg and was accomplished in more time (16). Symmetry of hip, knee, and ankle joints, angle during ambulation has also been studied in below-knee amputees, and it has been observed that the mean degree of symmetry of lower limb joints, angle was $0.802 \pm 0.04$ (9). During the loading response on early phase of weight acceptance, flexion of the knee under eccentric contraction of the quadriceps, provides a "shock absorption" effect which protects the joints from microdamage (1, 20).

Sequence of gait phases and symmetry of kinematic variables might be affected by various causes, which included deformities, muscle weakness, impaired motor control, and pain (4). In amputees, the following predominant causes are considered to interfere with normal gait:

1) Atrophy of and decrease in thigh muscle strength (12). A good control over the function of the stump-socket complex requires sufficient and effective strength of the muscles acting on the knee joint (14).
2) Stump length is the major factor determining stability of the BK stump inside the prosthetic socket and consequently the quality of gait. It was shown that by improving stump-socket stability,

length of stride becomes longer, speed of comfortable ambulation increases, and symmetry between stance phases improves (10).

3) Alignment effect on gait and standing sway. These have also been evaluated and investigators have attempted to establish the optimal range of alignment which is acceptable to the BK amputees (5, 11). Testing different alignment variables, it was found that foot alignment changes have the largest effect on gait symmetry (22).

4) The influence of the prosthesis itself on symmetry of gait movements was demonstrated by the existing differences between the amputee's normal and amputated limb. These differences were attributed mostly to the poor flexibility of the rigid ankle-foot component, which was found to slow the speed of dorsiflexion relative to normal and to fail in providing effective rapid plantarflexion at toe-off (1). Various "storing-energy" artificial feet have been developed in order to reproduce the function of a normal ankle and foot. Some were found to improve both the normal pattern of knee moments and the range of dorsal and plantar flexion of the artificial ankle and foot, thus returning some of the stored energy at push-off (3).

Assuming that a better understanding of processes altering normal gait events in BK amputees is likely to be of help in selecting the best technical means available for improving quality of gait, we therefore chose to investigate the periods of DLS and length of steps performed by the amputated and normal legs in BK amputees.

## MATERIALS AND METHODS

*Subjects*

We assessed thirteen volunteers (2 women and 11 men) with BK amputation, performed in 5 subjects on the right lower limb and in 8 subjects on the left limb. The ages of the subjects ranged from 27 to 65 years; mean age $42.6 \pm 14$ years, their heights from 163 to 186 cm; mean height $175.6 \pm 7.5$ cm, and their weights from 65 to 120 kg; mean weight $82.8 \pm 15.2$ kg. Eleven amputations were as a result of trauma, one from acute arterial thrombosis, and one due to peripheral arterial disease. The time lapse between the date of amputation and the time of fitting the first prosthesis ranged from one month to 18 months; mean time for receiving the first prosthesis was $5.07 \pm 4.2$ months. The time lapse between the date of amputation and the time of testing ranged from 3 years to 46 years, with a mean time lapse of $14.8 \pm 13.1$ years. Nine prostheses were patellar-tendon-bearing with suspension mechanism (belt - 5; thigh corset - 2). Four prostheses were patellar-tendon-supracondylar with an elastic sleeve for suspension. All prostheses had a solid-ankle-cushion-heel (SACH) feet.

All of the subjects were excellent walkers who used their prostheses on a regular basis and were leading an active normal family life. No subject was in permanent need of a walking aid. Some were using a cane when the stump was painful or when walking on unsafe terrain. A comfortable walking distance for the subjects ranged from 500 m to 10 km; mean $3.03 \pm 2.4$ km.

*Methods*

Before testing, all subjects were assessed by a prosthetist to ensure optimal fit and function of the prosthesis. None of the subjects had stump problems (blisters, sores, swelling or pain) on the testing day. All subjects were tested ambulating without the assistance of supporting aids.

Temporal and distance parameters were measured by means of a 10-m-long × 1-m-wide non-slip self-constructed conductive walkway with 10KΩ/m. Range of motion of the ankle, knee, and hip joints was measured by means of electrical goniometer (Penny & Giles Biometrics Limited, measuring range $\pm 180°$, infinite resolution). The working mechanism of the goniometers includes a series of strain gauges mounted around the inner circumference of a protective spring. These sensors were attached to the skin across the lateral aspect of the joints. Signals from the electrical contact system walkway and from the electrogoniometers were routed to an on-line computer and analysed. Subjects were instructed to ambulate at their most comfortable speed. Data were analysed using commercial software for computation and processing and for statistical analysis. The significance level of the differences was determined by using a non-parametric paired test (Wilcoxon signed ranks test). Results were judged to be statistically significant at $p < 0.05$.

## RESULTS

The average number of steps analysed in each subject was $20.9 \pm 3.6$, obtained during an uninterrupted walk along the conductive walkway. Means and standard deviations of gait cycle and stride distance, cadence and walking speed are summarized in Table I. Temporal and distance variables obtained in the amputated and in the normal legs and their degree of symmetry are compared in Table II. Differences between means and standard deviations of DLS time (amputated leg; $0.173 \pm 0.04$ seconds, normal leg; $0.211 \pm 0.05$ seconds) and step length (amputated leg;

Table I. *Means and standard deviations of gait variables obtained in the tested below-knee amputees ($n = 13$)*

| Gait variables | | Mean ± SD |
|---|---|---|
| Steps no. | | 20.9 ± 3.6 |
| Gait cycle | (s) | 1.08 ± 0.09 |
| Stride length | (m) | 1.461 ± 0.01 |
| Cadence | (steps/s) | 1.86 ± 0.05 |
| Speed | (m/s) | 1.34 ± 0.01 |

Table II. *Means and standard deviations of gait temporal and distance parameters obtained in both limbs and their relative symmetries*

| Gait parameters | | Amputated leg | Normal leg | $p$ | Symmetry |
|---|---|---|---|---|---|
| Stance | (s) | $0.725 \pm 0.06$ | $0.746 \pm 0.08$ | 0.055 | 0.92 |
| Swing | (s) | $0.359 \pm 0.05$ | $0.339 \pm 0.06$ | 0.157 | 0.85 |
| DLS | (s) | $0.211 \pm 0.05$ | $0.173 \pm 0.04$ | 0.011 | 0.74 |
| Step time | (s) | $0.551 \pm 0.05$ | $0.534 \pm 0.07$ | 0.3 | 0.86 |
| Step length | (m) | $0.752 \pm 0.08$ | $0.709 \pm 0.07$ | 0.045 | 0.90 |

Table III. *Hip and knee angles in both limbs obtained at different stages of double-limb support*

| Joint angle | Amputed leg | Normal leg | $p$ |
|---|---|---|---|
| Hip angle on heel-strike | $21.4 \pm 6.1$ | $19.5 \pm 4.1$ | 0.157 |
| Knee angle on loading response | $7.9 \pm 3.8$ | $14.4 \pm 4.1$ | 0.006 |
| Hip angle on toe-off | $3.4 \pm 3.7$ | $2.9 \pm 5.1$ | 0.3 |
| Knee angle on toe-off | $51.8 \pm 10.9$ | $44.6 \pm 11.1$ | 0.018 |

$0.752 \pm 0.08$ m, normal leg; $0.709 \pm 0.07$ m) were significant ($p = 0.045$).

Hip and knee angles relevant to DLS period were measured and compared in Table III. The following were considered: hip angle on heel-strike, angle of knee flexion on loading response which follows heel-strike, hip and knee angles on toe-off. Mean flexion of the knee measured during loading response of the amputated leg ($7.9 \pm 3.8°$) was significantly smaller ($p = 0.006$) than in the normal leg ($14.4 \pm 4.1°$). On toe-off, angle of knee flexion on the amputated leg ($51.8 \pm 10.9°$) was significantly greater ($p = 0.018$) when compared with the normal leg ($44.6 \pm 11.1°$).

## DISCUSSION

Twice during the period of stride, both limbs are supporting the body weight, once from heel-strike of the amputated leg until toe-off of the normal leg, and a second time from heel-strike of the normal leg until toe-off of the amputated leg (Fig. 1). Investigators have so far paid only limited attention to these periods

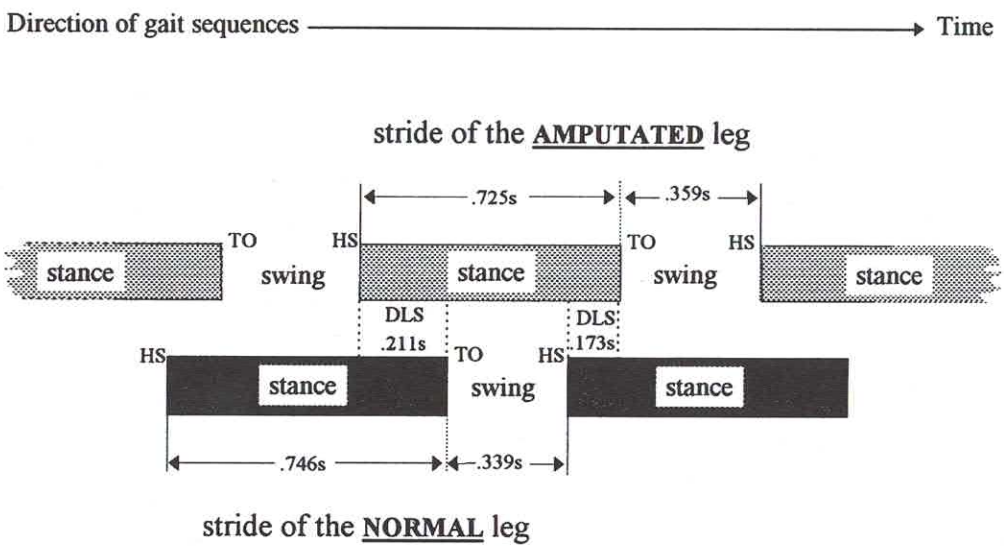


*Fig. 1.* Graphic description of double-limb support occurring during stride in below-knee amputees.

in amputees' gait. In the present work we obtained significant differences between periods of DLS ($p = 0.011$). We also obtained a symmetry of only 0.74 for double-limb support periods, which was the lowest degree of symmetry found among the gait parameters (stance, swing, step time and length). This asymmetry was due to a longer DLS time measured from heel-strike of the amputated leg until toe-off of the normal leg. We assume that this DLS is lengthened due to the rigid ankle-foot unit of most prostheses, which generates an internal dorsi flexion moment from heel contact to foot flat that rotates the prosthesis forward. This process is slow compared to the lowering of the normal foot under dorsi flexor control, and therefore the time of DLS is lengthened during forward motion of the amputated leg (1, 20).

We measured both hip joint angles at heel-strike, and found that differences were insignificant although hip flexion was greater in the normal leg. Since the loading response occurs during DLS, we compared the range of motion of both knees obtained at this phase. The mean flexion obtained in the amputated knee was significantly smaller ($p = 0.006$) indicating less effect of shock absorption during body weight acceptance. The event which terminates the period of DLS is toe-off of the leg facing the swing phase. Knee joint angles at toe-off were $51.8 \pm 10.9$ in the amputated leg and $44.6 \pm 11.1$ in the normal leg, a difference found to be significant ($p = 0.018$). We assume that the higher degree of flexion in the knee joint of the amputated leg results from the rigidity of the prosthetic ankle-foot unit. In fact, it has been noted (20) that no knee extensor moment is generated during prosthetic leg push-off, and therefore the energy transfer to raise the body's centre of gravity, so typical of normal gait, is lacking.

The role of the ankle plantar flexors during gait has been questioned by investigators, and published conclusions are controversial. It was observed that the plantar flexors contribute mainly to: (1) the stability of the knee and ankle, (2) the restraining of the forward rotation of the tibia on the talus during stance phase, and (3) the conservation of energy by minimizing vertical oscillation of the whole body centre of mass. Also, the plantar flexors do not propel the body forward. Paradoxically, maximum step length is not possible without the stabilizing effect of these muscles (18). Nevertheless, others found that both the soleus and medial gastrocnemius reach peak activity at terminal stance, during the high-powered push-off period when the ankle undergoes rapid plantar flexion, indicating a rapid generation of energy to propel the limb upward and forward (19, 21). In amputees, the contracting hip extensors of the amputated leg, generate energy to propel the body forward, thus compensating for the lack of ankle plantar flexor activity (20). Our results indicate a significant discrepancy in length of steps among our subjects. We assume that this difference is directly related to the rigid prosthetic ankle-foot unit which lacks the active, rapid plantar flexion burst at push-off. These results have been found by others (20). It is therefore recommended that when discussing step-length differences in BK amputees it might be more accurate to state that an amputee takes a shorter step with the normal limb (the "pathological" step) rather than a longer step with the amputated limb (the normal length step) (8, 15).

## ACKNOWLEDGEMENT

The authors extend their thanks to Bevetek Tomaz, physical therapist, for his assistance during the testing procedures.

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