Equilibrium and movement control strategies in trans-tibial amputees


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Abstract
This study was aimed at identifying changes in equilibrium and movement control strategies in trans-tibial amputees (TTA) related to both the biomechanical changes and the loss of afferent inflow. The coordinations between equilibrium and movement were studied in traumatical TTA and in controls during transition from bipedal to monopodal stance. TTA failed to perform the task in a high percentage of trials both when the sound and the prosthetic limb were supporting. Significant differences were also found between TTA and controls in the duration of the weight transfer phase, in the length of the initial centre of pressure (CP) displacement and in the electromyographic (EMG) patterns. Despite adaptive posturomotor control strategies, transition from bipedal to monopodal stance remains a difficult task to perform for TTA, both when the supporting limb is the affected one and when the sound one is. The results of this study are discussed with respect to the rehabilitation programme and the prosthesis design for trans-tibial amputees.

Introduction
Trans-tibial amputation is responsible for biomechanical changes (i.e. muscles, bones and joints missing) and for modifications in both afferent and efferent projections. Because of these impairments, equilibrium is difficult to control and falls are a significant problem for trans-tibial amputees (TTA) (Vlahov et al., 1990).

Several studies have shown that standing balance was altered in lower limb amputees. Fernie and Holliday (1978), Isakov et al. (1992), Geurst et al. (1991) found increased sway in lower limb amputees as compared to controls. In the study by Hermodsson et al. (1994), in patients with amputation of traumatic or vascular origin and in controls, lateral sway was increased in amputees as compared to controls, thus indicating a reduced balance capacity. By contrast, Vittas et al. (1986) found a decreased sway in patients with trans-tibial amputation.

These studies have shown that balance control strategies of amputees differed significantly from those of able-bodied subjects. In accordance with these findings, rehabilitation programmes of TTA include specific training aimed at improving balance during bipedal stance. It is implicit in this approach that improved balance during bipedal stance should result in better locomotor performances (Rodgers et al., 1993).

However, measurements of static standing balance do not necessarily characterise the balance performances during movement activities involving the lower limb such as transition from bipedal to monopodal stance or walking. Hence, rehabilitation approaches, which only rely on a static analysis of balance,
may not be suited for movements in which body segments are undergoing substantial changes in acceleration. Thus the elaboration of rehabilitation programme for the restoration of upright mobility in TTA may be improved through a better knowledge of the kinesiological factors underlying the execution of motor tasks involving the lower limbs, such as lateral leg raising (Mouchino et al., 1992) or single leg flexion (Rogers and Paï, 1990), or gait initiation (Brenière et al., 1981). These tasks, which imply a transition from bipedal to monopodal stance, appear to be difficult for amputees, and to increase the risks of falls (Vlahov et al., 1990).

The analysis of kinetic, kinematic and EMG recordings data has provided information about the coordination between posture and movement during single leg flexion (Rogers and Paï, 1990), and lateral leg raising (Mouchino et al., 1992), which have shown to be complex biomechanical tasks. In those tasks involving a limb that has previously been supporting part of body weight, movement onset is preceded by postural adjustments (Brenière et al., 1981; Brenière and Do, 1991; Massion, 1994) serving to shift the centre of gravity (CG) towards the supporting leg so that the moving leg can be raised and equilibrium maintained during the movement. It was shown in previous studies (Mouchino et al., 1992) that a horizontal force was exerted on the ground by the leg to be moved, which was shortly preceded by the activation of the gastrocnemius medialis (GM).

Trans-tibial amputees are deprived of active distal muscles including the GM. Furthermore, despite their prosthesis, amputees do not recover their afferent inflow (i.e. proprioceptors residing in the amputated portion of the leg) or their degrees-of-freedom of the lower limb. The purpose of the authors, study was to identify changes in equilibrium and movement control strategies in TTA during transition from bipedal to monopodal stance in order to improve both the assessment of balance function and the management of the rehabilitation programme of these patients.

Materials and methods

Population

The test subjects included 5 patients with unilateral transfemoral amputation of traumatic origin (5 males, mean age 34.8 years, range 24 to 49). As regards the surgical procedure, myoplasty was undertaken in all cases. They did not present signs of locomotor, neurological, vascular or other pathology that could influence stance and gait. The patients used the same type of prosthesis including a total contact socket and a SACH foot.

Control subjects were 5 age-matched healthy males.

Materials

The kinematic study was carried out using an optoelectronic system (ELITE, Bioengineering Technology and Systems, via Capecelatro 66, 20148, Milano, Italy). Twelve (12) light-reflecting markers were placed on anatomical landmarks: bilaterally, on the infra orbital margins, acromions, anterior-superior iliac crests, greater trochanters, medial border of the tibial plateau, medial malleoli. The movements of the markers were recorded at a frequency of 100Hz by four infra-red cameras, positioned in front of the subjects.

The kinetic parameters (ground reaction forces) were recorded from a Kistler force plate (Kistler Instrument Corp. 75 John Glenn Drive, Amherst, New York, 14120, USA).

Electromyographic (EMG) recordings were performed on 8 muscles by means of surface electrodes. The EMG signals were amplified, filtered, digitized at 500Hz and rectified. The following muscles were studied: gluteus medius (Glut), tensor fasciae latae (TFL), vastus lateralis (VL) on both sides, tibialis anterior (TA) and gastrocnemius medialis (GM) on the sound limb of the amputees and on the moving leg of the control subjects.

Experimental procedure

Each subject was asked to stand on the force-plate with his feet 100mm apart, his hands behind his back and his eyes gazing at two electroluminescent diodes placed symmetrically 5m in front of the subject’s eyes. In response to a light indicating the side on which the movement had to be performed, the standing subjects were instructed to raise laterally one leg as fast as possible to a 45° angle relative to the vertical axis and to maintain the final position for a few seconds. Twenty trials were performed in a randomised order. Recordings were performed for 3000ms, the onset of the diode light occurring 300ms after the beginning of the recordings.
Parameters and data analysis

Trials were defined as either balanced or imbalanced. Imbalanced trials were those in which the CG was not maintained for at least 2s in the same place and/or in which the final leg and trunk positions were not stabilised for at least 2s.

The velocity of the marker placed on the moving malleolus.

The GG position was calculated using a model, the displacements of the markers placed on the anatomical landmarks determined using the movement analysis system and anthropometric values.

The displacement of the centre of pressure (CP) was determined from the force plate recordings. The CP is the point of application of the resultant of the vertical ground reaction forces. The amplitude of the CP displacement was calculated from the initial CP position. The value of the regression line based on the CP displacement in the frontal plane, starting from the time tbal (end of the ballistic component of the GG transfer) to the tbal + 1000ms, was computed for each trial and from this value the root mean square (RMS) of the CP displacement data was calculated.

Using kinetic and kinematic parameters, three consecutive phases were identified during the leg movement (Mouchnino et al., 1992). The "transfer phase" of the body weight towards the supporting side begins with the onset of the first CP displacement in the frontal plane (t1) and ends up with the onset of the change of the vertical velocity of the moving malleolus marker (t2). A "movement phase" follows until the end of the leg raising, when the vertical velocity of the malleolus marker returns to 0 (t3). The third phase is the "position maintenance phase" which continues up to the end of the recordings.

EMG analysis was performed by calculating the latencies and areas of EMG activities or bursts. The resting activities were measured in each trial during 300ms preceding the starting signal in order to determine the background EMG activity. The mean and standard deviation of this background activity were then calculated for each subject under each condition. Timing and intensity measurements were performed. For the timing measurements, the onsets and offsets of the EMG bursts were defined as the times at which the EMG activity increased or decreased below a threshold level set at twice the standard deviation of the background activity above baseline level. The duration of the burst was also calculated. The intensity of the muscular activity was calculated by subtracting the baseline from the EMG activity level reached 300ms after the activity had increased above the threshold level.

Statistical analysis was performed on all data (n=200) recorded from the individuals. Analysis of variance (ANOVAs) was used to analyse the data with the group of participants (control and amputees) as a between-participants variable and sound versus prosthetic leg in amputees as a within-participants variable. Student's t-test was used for paired variables and comparisons between conditions. The level of significance was taken to be at least p<0.05 level.

Results

Balanced and imbalanced trials

In the subjects, 30% of the trials were imbalanced when the sound leg was the supporting one (s.d.7) and 32% of the trials were imbalanced when the prosthetic limb was the supporting one (s.d.4). Controls failed to perform the task in only 4% of the trials (s.d.5).

Centre of gravity position

To investigate the subjects’ CG position, the CG vertical projection with respect to the ankle-joint axis was analysed. Before the starting signal, when the subject was in a standing position, the CG position in the frontal plane was located close to the medial line (half of the inter-malleolus distance) in the controls (2.4mm+/−0.76) as well as in the amputees (6.97mm+/−1.04), where it was located slightly nearer the prosthetic leg. The initial CG position in the sagittal plane was located in front of the ankle joint axis, and was slightly further backward (12.8mm+/−2.7) in amputees than in controls.

The CG shift in the frontal plane was larger in controls (131mm+/−30) than in amputees (114mm+/−21), (F(1.146)=15.54; p<0.0001). In amputees, no effect of side was noted (F(1.91)=0.112; p=0.74). Differences in the CG shift were, however, observed between balanced and imbalanced trials when the sound leg was the supporting one (F(1.51)=23.07; p<0.0001). In that case, the CG shift reached 112mm+/−2.1 in balanced trials and 54mm+/−6 in the
imbalanced trials. No difference was found in the sagittal plane. When the prosthetic leg was supporting, no difference in the CG shift in the frontal plane was observed between balanced (104mm+/−35) and imbalanced trials (115mm+/−20) (F(1.48)=0.35; p=0.56).

Centre of pressure stabilisation

The RMS (Fig. 1) was calculated to quantify the CP stabilisation on one foot. In a general way, the results indicated that there was a slight population effect (amputees vs controls) on the RMS (F(1.208)=5.24; p<0.02). The RMS was higher in the controls (5.54mm+/−2.78) than in the amputees (4.69+/−2.55). In the amputees, there was a side-effect (prosthetic vs sound moving leg) on the RMS (F(1.99)=21.28; p<0.001). When the sound leg was supporting

![Diagram showing RMS displacement values averaged across the subjects.](image-url)
the body, the RMS was 5.93mm (+/-2.48) and was equal to that observed in the controls (5.54mm+/2.78). When the prosthetic leg was supporting the body, the RMS value was the lowest (3.76mm+/2.19).

**Phase durations**

The weight transfer phase was longer in amputees (467ms+/-168) than in control subjects (371ms+/-153), (F(1.209)=24.7; p<0.0001). No effect of side was found in the amputees (F(l,99)=2.79; p<0.09).

The movement phase was also longer in amputees (1265ms+/-277) than in control subjects (1049ms+/-230) (F(1.196)=35.5; p<0.0001). No effect of side was observed in the amputees (F(l,97)=0.17; p=0.68).

**Centre of pressure displacement**

The amplitude of the first peak of the CP displacement, which is assumed to constitute the initiation of the body weight transfer (9), was higher in control subjects (99mm+/-39) than in amputees (sound leg raised: 80mm+/-34; prosthetic leg raised: 85mm+/-30) no matter which leg was the moving one (F(1.209)=17.21; p<0.0001). No effect of side (F(1.91)=1.14; p=0.28) or of type of trials (balanced vs imbalanced) (F(1.91)=0.27; p=0.60) was noticed.

**EMG activities during body weight transfer**

(Figs. 2 and 3)

When the sound leg was about to be raised, the early classical GM activation was time-locked with t1 in both the amputees (27ms+/-110) and the controls (-15ms+/-71). No significant differences were observed between the controls and the amputees in either the intensity or the duration of the GM burst. The GM burst lasted 400ms (+/-214) in the amputees and 342ms (+/-183) in the controls, and no group effect (amputees vs controls) was observed (F(1.107)=0.653; p=0.42). In amputees, however, the GM is coactive with the TA, which was delayed by 33ms (+/-123). It was also observed that a TFL burst was present in 68% of the trials performed with the sound leg and was time-locked with the GM burst. No such TA and TFL co-activation was ever observed in the controls at the time.

When the prosthetic leg was about to be raised, only one muscle, the TFL, showed an early burst, which was time-locked (34ms+/-2.27) with the onset of the CP change (t1) and might be involved in initiating the CP thrust. Its onset did not occur systematically before t1, however. The duration of the TFL burst (285ms+/-105) was statistically equivalent to the CP shift duration defined between t1 and the instant of the peak (274ms+/-76). No comparable phasic activation was observed in the TFL in the control group. The TFL burst was shorter (200ms+/-126) when the sound leg was raised than when it was the prosthetic one (F(1,66)=12.94; p<0.005).

**Discussion**

Previous studies on equilibrium in lower limb amputees have focused on the analysis of balance during bipedal stance. In a number of these works (Fernie and Holliday, 1978; Geurst et al., 1991; Isakov et al., 1992; Vittas et al., 1986), patients with amputation of various origin (i.e. vascular, traumatic, infection) and of various levels (i.e. trans-tibial and trans-femoral) were analysed together. The lack of selective criteria can probably explain the contradictory results recorded in these studies. Hence, Hermodsson et al. (1994) demonstrated that postural sway was markedly different in patients with amputation of traumatic origin and in patients with amputation of vascular origin. Furthermore, Hermodsson et al. (1994) and Vittas et al. (1986), also showed differences in postural behaviour depending on the age of the patients. In accordance with these findings, the patients included in the present study, although of limited number, were comparable with respect to the origin of the amputation (i.e. trauma), the level of amputation (trans-tibial) and the age (under 50 years old).

When comparing postural sway during bipedal standing, in TTA and controls, Hermodsson et al. (1994) showed that traumatic amputees exhibited mild differences as compared to healthy subjects. Only, the sagittal sway was found to be decreased in amputees as compared to healthy subjects. In the present study, the changes in equilibrium and movement control strategies were assessed during transition from bipedal to monopodal stance (i.e. lateral leg raising), using kinematic, kinetic and EMG data. The displacement of the centre of gravity, the stabilisation of the centre of pressure and the initial CP shift as well as the phases durations
Fig. 2. Kinetic and rectified EMG patterns obtained with one amputee.

A: The moving leg was the prosthetic one, with tibialis anterior (TA) and gastrocnemius medialis (GM) lacking.

B: The moving leg was the sound one. The times plotted on this curve were t1 (onset of the centre of pressure (CP) change) and t2 (onset of the movement measured on the vertical velocity curve). Note that on both sides, the tensor fasciae latae (TFL) was activated at t1 and could be involved in the thrust. Note the TA and GM co-activation occurring during the thrust.
Fig. 3. Rectified EMG patterns recorded from one control subject. The EMG were recorded at a proximal level (TFL tensor fasciae latae, Glut. gluteus medius) and at a distal level (TA tibialis anterior, GM gastrocnemius medialis). Note the GM and TA burst sequence.
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and the muscle activity patterns were selectively modified in trans-tibial amputees as compared to control subjects. Therefore, the study of lateral leg raising provides additional information to the analysis of postural sway during static standing for the assessment of balance function in TTA.

Moreover, Hermodsson et al. (1994) found that balance during one leg-standing on the sound side was not altered in TTA as compared to controls. On the other hand, in the same study, only one patient succeeded in standing on the prosthetic leg. In the authors' work, a high percentage of imbalanced trials (30%), was recorded during lateral leg raising both when the sound leg was supporting and when the prosthetic leg was supporting. Thus the study of transition from bipedal to monopodal stance indicates significant alteration in equilibrium and movement control in TTA even when the sound leg is the supporting one.

The analysis of the imbalanced trials provides information for the elaboration of the rehabilitation programme as well as for the prosthetic design. When the sound leg is supporting, the same percentage of failures is recorded as when the prosthetic limb is supporting. This result, which might not have been expected, is explained by a reduced displacement of the centre of gravity which was not shifted far enough laterally to be located within the supporting foot area. The GM activity of the leg to be raised, which admittedly initiates the CP mediolateral thrust (Mouchnino et al., 1992; Rodgers et al., 1993) onto the ground in normal subjects, is lacking in a trans-tibial amputated leg. How the CP mediolateral thrust is initiated in that case be explained by the burst of proximal muscles. Hence, an unusual activity of the TFL on the moving side, time-locked with the onset of the CP shift was recorded in amputees (Fig. 2), when the prosthetic leg was to be raised. Thus, the TFL may be one of the muscles responsible for the initiation of the mediolateral thrust at a proximal level when that leg is still on the ground. Therefore, the recordings of muscles' activities during lateral leg raising in trans-tibial amputees show that muscles such as the TFL which do not usually take part in the leg raising process are involved in the onset of the movement. This suggests that muscles other than those tested here may also have been involved in initiating the CP mediolateral thrust. These findings suggest inclusion in the rehabilitation programme of trans-tibial amputees of a specific training of proximal muscles such as the TFL on the side of the amputation during transition from bipedal to monopodal stance. However, further studies will be necessary to determine which other muscles are involved in balance during lateral leg raising in amputees.

When the prosthetic leg was the supporting one, the root mean square (RMS), calculated to quantify the CP stabilisation on one foot, proved to be lower than when the sound leg was supporting and than in the control subjects (Fig. 1). Since it is admitted that ankle control plays a major role in mediolateral balance (Winter et al., 1996), the results of imbalance trials can be explained by the reduced capacity to stabilise the position of the centre of pressure on the prosthetic leg, related to the limited ankle joint mobility on the prosthetic side and to the lack of distal active muscles. Furthermore, trans-tibial amputees have to deal with the loss of the afferent inflow mainly from proprioceptors residing in the amputated portion of the leg and foot plantar cutaneous receptors. Hence, amputees cannot use proprioceptive information or distal muscles to realise corrective adjustments and have to reach the stabilisation point of their centre of pressure straight away. Thus, when performing lateral leg raising with the prosthetic leg supporting, equilibrium will be found if the presetting of the centre of pressure position has been accurate. By contrast, if adjustments of the CP position are needed, the amputee will not be able to perform the task. A previous study (Mouchnino et al., 1992) has shown that dancers developed a similar postural mechanism so as to reach their equilibrium point without subsequent CP adjustments. The accuracy of the presetting of the CP control reflected the level of training. These findings suggest that TTA could improve the CP control through specific rehabilitation techniques, focusing on transition from bipedal to monopodal stance. Moreover, techniques such as CP biofeedback (Shumway-Cook et al., 1988), which have previously been developed in hemiparetic patients, could be helpful to improve the CP control in trans-tibial amputees. Finally, it can also be suggested that, since they develop adaptive posturomotor strategies using a prosthesis, trans-tibial amputees should be given the opportunity to use a prosthesis as soon as
possible after amputation in order to achieve better results.

The results of the present study also provide some information for the prosthesis design. The loss of afferent inflow from proprioceptors residing in the amputated portion of the leg and from foot plantar cutaneous receptors in trans-tibial amputees is partly responsible for the reduced capacity to stabilise the centre of pressure in the frontal plane. It can be hypothesised that, as stated by Goldberg and Mayer (1996), the skin of the stump which becomes more sensitive to pressure at the skin socket interface, facilitates the prosthesis movement control. It is, therefore, likely that prosthesis design and especially total contact sockets should improve the afferent inflow from the stump and help to obtain a better coordination between equilibrium and movement in trans-tibial amputees. Besides, despite their prostheses, amputees do not recover their degrees-of-freedom of the lower limb. For this reason, some prostheses have been designed in order to provide an increased mobility in the frontal plane at the ankle level. However, TTA show a reduced capacity to stabilise the centre of pressure in the frontal plane, as shown by the reduced value of the root mean square of the CP displacement (Fig. 1). It can, therefore, be hypothesised that because of the lack of active distal muscles, the increased prosthesis mobility in the frontal plane will make equilibrium control even more difficult to achieve.

Conclusion

This preliminary report has shown that the study of lateral leg raising brings up additional information to standing posture analysis for the assessment of balance function in traumatic trans-tibial amputees. These results provide the means of a new approach to the development of the rehabilitation programme and to prosthetic design for trans-tibial amputees. Further study will concern a larger number of trans-tibial amputees before and after undergoing equilibrium and movement control oriented rehabilitation programmes.

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