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Dynamic Balance Control (DBC) in lower leg amputee subjects; contribution of the regulatory activity of the prosthesis side

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ABSTRACT

Background: Regaining effective postural control after lower limb amputation requires complex adaptation strategies in both the prosthesis side and the non-amputated side. The objective in this study is to determine the individual contribution of the ankle torques generated by both legs in balance control during dynamic conditions.

Methods: Subjects (6 transfemoral and 8 transtibial amputees) stood on a force platform mounted on a motion platform and were instructed to stand quietly. The experiment consisted of 1 static and 3 perturbation trials of 90 s duration each. The perturbation trials consisted of continuous randomized sinusoidal platform movements of different amplitude in the sagittal plane. Weight distribution during the static and dynamic perturbation trials was calculated by dividing the average vertical force below the prosthesis foot by the sum of forces below both feet. The Dynamic Balance Control represents the ratio between the stabilizing mechanism of the prosthetic leg and the stabilizing mechanism of the non-amputated leg. The stabilizing mechanism is calculated from the corrective ankle torque in response to sway. The relationship between the prosthetic ankle stiffness and the performance during the platform perturbations was calculated.

Findings: All patients showed a (non-significant) weight bearing asymmetry in favor of the non-amputated leg. The Dynamic Balance Control ratio showed that the contribution of both legs to balance control was even more asymmetrical. Moreover, the actual balance contribution of each leg was not tightly coupled to weight bearing in each leg, as was the case in healthy controls. There was a significant positive correlation between the prosthetic ankle stiffness and the Dynamic Balance Control.

Interpretation: The Dynamic Balance Control provides, in addition to weight distribution, information to what extent the stabilizing mechanism of the corrective ankle torque of both legs contributes to balance control. Knowledge of the stiffness properties may optimize the prescription process of prosthetic foot in lower leg amputee subjects in relation to standing stability.

1. Introduction

One of the main goals during the rehabilitation of amputee patients is regaining balance control. Traditional balance training is aimed at equal loading on both legs during standing and walking (Esquenazi and DiGiacomo, 2001). As such, the success of this approach is evaluated in terms of static or dynamic weight distribution between the legs, determined by descriptive measures obtained from the position or movement of the Centre of Pressure (CoP) (Aruin et al., 1997; Buckley et al., 2002; Hermodsson et al., 1994; Isakov et al., 1992; Summers et al., 1987; Vittas et al., 1986). Several longitudinal studies demonstrate that in amputees weight bearing symmetry is regained within 8 weeks after receiving the first prosthesis (Geurts et al., 1991; Stolov et al., 1971), however, in many cases there is still further functional improvement beyond this period. This suggests that symmetrical weight distribution does not fully reflect balance restoration. The central program that is responsible for the maintenance of balance consists of two main strategies, namely, ankle strategy and (proximal) hip strategy (Horak and Nashner, 1986). Obviously, in lower leg amputees, the ability to utilize the ankle strategy is absent, however, the stiffness of the prosthetic ankle/foot complex (or Prosthetic Ankle Stiffness = PAS) can enable an amputee to create ankle torque. The ankle torque is an inherent property of the prosthetic foot and influences the stance phase biomechanics of gait (Gitter et al., 1991; Prinsen et al., 2011). As such, it can be expected that the stiffness characteristics of the prosthetic feet also influence performance on standing balance.

In the present study, we were interested whether further functional improvement after achieving weight symmetry is the result of optimizing the utilization of the PAS, or results from compensatory changes in the intact non-amputated side (Aruin et al., 1997; Czerniecki, 1996; Seroussi et al., 1996; Vrieling et al., 2008).

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In addition, we were interested in the role of the stiffness characteristics of the prosthetic feet in the control of balance.

The common descriptive measures obtained from the Centre of Pressure (CoP) trajectory or the ground reaction forces do not sufficiently quantify the actual contribution of the PAS to balance control. For this purpose we need to take into account, not only the generated regulatory activity of both legs but also relate it to the balance performance. Dynamic balance contribution (DBC) is a direct reflection of the generated ankle torque in reaction to (external) perturbations of the centre of mass (CoM) (Simons et al., 2009; van Asseldonk et al., 2006) and provides a way to disentangle the contribution of the individual ankle joints to balance control. In non-amputees, the DBC indicates the combined stabilizing effect of all muscle and connective tissues crossing the ankle. In amputees, the DBC reflects the utilization of the PAS in balance control.

The objective of this study was to determine whether the DBC contains additional information besides the weight distribution that can explain further functional improvement beyond the 8 weeks period during which weight bearing symmetry has been achieved. The primary goal was to ascertain the contribution of both legs in balance control individually in experienced lower limb prosthetic users (transfemoral and transtibial) by assessing the generated ankle torque. The secondary goal was to determine whether the contribution of the prosthetic leg is related to weight bearing of the prosthetic leg. The third goal was to determine whether the contribution of prosthetic ankle in balance control is related to the stiffness of the prosthetic ankle (PAS).

2. Methods

2.1. Subjects

Six transfemoral and 8 transtibial unilateral amputee patients participated in the study (Table 1). Both groups were experienced lower limb prosthetic users. There were various reasons for amputations, such as trauma, malignancy, vascular insufficiency. Subjects' functional level varied from limited outdoor to active walker with the ability to walk at slow or fast speeds and negotiate small obstacles. Subjects were able to walk with variable cadence (approximately K-3 according to the Medicare Functional Classification). Subjects were excluded if they suffered from other musculoskeletal or neurological problems influencing their balance or walking ability such as vascular disorders or peripheral neuropathy; or they had stump problems/bad socket fitting. Transfemoral amputee patients used different mono and polycentric unlocked knees with free swing. Laser Assisted Static Alignment Reference (LASAR posture device; Otto Bock Health Care North America, Minneapolis, MN, USA) was used for optimal alignment in both groups. With this method, the vertical component of the ground reaction force is visualized by a laser line projected from a force plate on the subject's leg. By adjusting the sagittal foot position, this line is positioned in relation to the knee axis. For the transtibial amputees the laser line was positioned through the anatomical knee axis. In the transfemoral amputees wearing the C-leg and the 7 axis knee the laser line was positioned 0–5 mm in front of the knee axis and in other subjects it was positioned more than 5 mm in front of the knee axis. The results of earlier similar experiments in a group of 6 healthy control subjects (van Asseldonk et al., 2006) were used for reference values.

The study conformed to the principles of the Declaration of Helsinki and was approved by the local medical ethics committee. All subjects gave their informed written consent prior to start of the experiment.

3. Experimental set-up

3.1. Procedures and equipment

Body height and weight of every subject were measured before the start of the experiment. The subjects stood on a forceplate and faced a light gray background. Their feet were placed with a 20 cm distance between the medial malleoli and 9° outward rotation with respect to the sagittal midline.

The forceplate was embedded within a computer-controlled dynamic motion platform with six degrees of freedom. The custom made forceplate consisted of four force sensors (ATI-Mini 45SI-580-20) with six degrees of freedom, mounted in a rectangular configuration on an aluminum plate. Each sensor was covered with a rectangular, 15×17.5 cm, aluminum plate. A foot frame ensured that each foot was placed solely on the cover of two force sensors. Forces and torques measured by each sensor were sampled at 360 Hz.

Reflective spherical markers were attached to the heel, big toe, lateral malleolus and knee, and halfway the lateral tibia and thigh of both legs, as well as to the sacrum, head and shoulders. In addition, a cluster of 3 markers was attached on both anterior superior iliac spines and 3 markers were attached to the platform. The positions of the markers were recorded at 120 Hz using a three-dimensional motion analysis system consisting six video cameras and a control unit (VICON).

At the start of the experiment, during a static trial, the subjects were instructed to stand quietly in the anatomical position for 10 s. Next, during 3 dynamic trials of 90 s duration, they were instructed to ‘maintain their balance without moving their feet’ while random

Table 1
Characteristics of the amputation groups.

Subject	Level	Age	Sex	Weight (kg)	Length (cm)	Years since amputation	Cause	Side	Foot	Knee	Suspension	PAS Nm/Rad	Acceleration of platform perturbations	
													Max	Min
1	Tf	57	♂	60	164	6	M	r	Dynamic	Mpc	Liner	521	1.02	−0.97
2	Tf	52	♂	90	170	16	M	l	1-axis	4-axis	Suction	604	0.72	−0.072
3	Tf	49	♀	78	170	13	V	r	Dynamic	4-axis	Suction	235	0.70	−0.070
4	Tf	63	♂	79	180	12	T	r	1-axis	4-axis	Liner	276	0.74	−0.76
5	Tf	45	♂	66	184	8	T	l	Dynamic	7-axis	Liner	475	0.82	−0.84
6	Tf	48	♂	88	178	7	T	l	Dynamic	Hydraulic 1-axis	Liner	395	1.37	−1.38
7	Tt	60	♂	84	183	4	V	l	Dynamic		Liner	162	1.08	−1.04
8	Tt	62	♂	101	194	20	T	l	Dynamic		Liner	704	0.95	−0.98
9	Tt	61	♂	81	189	10	T	l	Dynamic		Liner	372	0.70	−0.72
10	Tt	36	♂	105	190	11	T	l	Dynamic		Liner	384	1.12	−1.10
11	Tt	38	♂	100	186	1	T	l	Dynamic		Liner	355	1.23	−1.23
12	Tt	33	♂	76	179	5	T	l	Dynamic		kbm	428	0.91	−0.97
13	Tt	68	♂	100	178	11	T	r	Dynamic		Liner	286	0.94	−0.99
14	Tt	35	♀	60	173	32	C	r	Dynamic		Liner	186	0.94	−1.00

Tf = transfemoral amputation, Tt = transtibial amputation, r = right, l = left, T = trauma, V = vascular, M = malignancy, C = congenital, mpc = microprocessor controlled knee, PAS = prosthetic ankle stiffness.