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Local Dynamic Stability of Lower Extremity Joints in Lower Limb Amputees during Slope Walking

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Abstract—**Lower limb amputees have a higher fall risk during slope walking compared with non-amputees. However, studies on amputees' slope walking were not well addressed. The aim of this study was to identify the difference of slope walking between amputees and non-amputees. Lyapunov exponents λ_S was used to estimate the local dynamic stability of 7 transtibial amputees' and 7 controls' lower extremity joint kinematics during uphill and downhill walking. Compared with the controls, amputees exhibited significantly lower λ_S in hip (P=0.04) and ankle (P=0.01) joints of the sound limb, and hip joints (P=0.01) of the prosthetic limb during uphill walking, while they exhibited significantly lower λ_S in knee (P=0.02) and ankle (P=0.03) joints of the sound limb, and hip joints (P=0.03) of the prosthetic limb during downhill walking. Compared with amputees level walking, they exhibited significantly lower λ_S in ankle joints of the sound limb during both uphill (P=0.01) and downhill walking (P=0.01). We hypothesized that the better local dynamic stability of amputees was caused by compensation strategy during slope walking.**

I. Introduction

Amputees have a higher fall risk compared with non-amputees. Previous study showed that 16.5% amputees fell at least once in the surgical ward during post-operative recovery period, and injuries were sustained in 60.7% of those who fell [1]. And amputees are specially challenged by walking in complex environment, such as irregular surface, stair and slope. Former studies of amputee gait focused on level walking, stair walking, obstacle crossing and turning [2-5]. However, studies on amputees' slope walking were not well addressed.

Slope walking is quite common in our daily activities, and has a greater fall risk than level walking and stair walking [6]. Previous studies found that amputees had different gait velocity and lower limb joint angles compared with the non-amputees [7]. In these studies, postural sway and gait parameters were frequently used to evaluate amputees' motion stability [8-10]. However, the mechanisms governing standing and walking stability are significantly different [11]. Furthermore, analyzing isolated and independent strides in typical gait researches is not quite suitable to assess how human maintain dynamic stability during walking.

Recently, there is a method of quantifying human walking local dynamic stability by estimating maximum finite-time Lyapunov exponents (λ_S: short-term exponents computed between 0 and 1 stride, and λ_L: long-term exponents computed between 4 and 10 strides) during continuous treadmill walking [12]. It characterizes how human walking system responds to perturbations. If $\lambda>0$, it indicates the system is unstable, otherwise the system is stable. The higher the λ is the more unstable the system is. It was found that the local dynamic stability of anterior-posterior and medio-lateral trunk accelerations of lower limb amputees were poorer than the controls during multi-condition level walking [13]. But researches on amputee lower joints dynamic stability during slope walking were not well addressed so far.

The purpose of this study was to identify the difference of slope walking stability between amputees and non-amputees. Lyapunov exponents of lower limb joints' kinematics were calculated to estimate walking dynamic stability. Therefore, we can have an insight into amputees' slope walking and assist the prosthetic design and rehabilitation courses in future.

II. Method

A. Subjects

Amputee volunteers were approached via prosthetic workshops and rehabilitation centers. Inclusion criteria: (1) adults with unilateral transtibial amputation at least one year; (2) adaption to daily use of prosthetic; (3) ability of walking more than 50m without aids; (4) no history of falls for the last 6 months; (5) being free from any medical conditions that affect them to complete the experiment, such as neurological and orthopedic disorders, severe visual impairment, and stump pain. A control group of non-amputees were recruited from the local students and staffs.

7 unilateral transtibial amputee subjects (5 male, 2 female, age 40.3(6.7) years, years since amputation 5.0(3.7) years, height 166.9(8.2) cm, weight 66.3(12.4) kg, normal walking speed 1.4 (0.32) km/h) and 7 non-amputee subjects (5 male, 2 female, age 35(10.5) years, height 169.4(8.2) cm, weight 65.1(9.9) kg, normal walking speed 2.3(0.42) km/h) participated the experiment. The study was approved by the ethical committee of School of Biomedical Engineering Shanghai Jiao Tong University. All subjects signed informed consent before testing.

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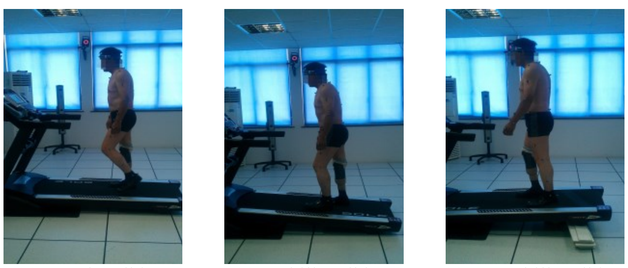
B. Procedure

The study was performed in a motion analysis laboratory. A Vicon™ T40 system and Nexus suite (Oxford Metrics, Oxford, UK) were used to capture and preprocess the subjects' motion. 39 reflective markers (14-mm spheres) were placed on each subject at bony landmarks according to Vicon's Plug-In-Gait full-body (UPA and FRM) model. A treadmill (Sole, USA) was used to acquire continuous strides for analysis.

Subjects first walked on the treadmill for 3 minutes with a self-selected pace to adjust themselves to the testing and identify their comfortable walking speed. After the preliminary trial, subjects were instructed to walk on the treadmill under three conditions: level, 10° uphill and 10° downhill (Figure 1). The previous research pointed out that non-amputees' lower joint kinematics varied with inclines from -10° to 10° [14]. According to our preliminary experiments of non-amputees' walking on 3°,6°, 9° and 12° uphill and downhill treadmill, significant lower joint kinematical differences were observed around 9° uphill and downhill walking. Thus, the slope inclined angles were set as 10° uphill and downhill in this study. Under each condition, subjects completed a trial of continuous walking lasting for at least 5 minutes. Between each trial, subjects were asked to rest for a few minutes.



(c) Level walking (c) Uphill walking (c) Downhill walking

Figure 1. Experiment design. *((a)amputees walked on a level treadmill (b)amputees walked on a 10° up-inclined treadmill (c) amputees walked on a 10° down-inclined treadmill)*

30 consecutive strides were picked out from raw data series, and resampled. Lower joint angles of F/E (flexion/extension), IR/ER (internal rotation/external rotation), and Abd/Add (abduction/adduction) were calculated by Vicon Nexus. Lyapunov exponents of three rotational angles of hip, knee, and ankle joints were calculated to estimate the local dynamic stability of subjects' walking.

C. Calculation of Lyapunov exponents

In the present study, local dynamic stability was estimated by Rosenstein's algorithm [15]. In this algorithm, an appropriate state space should be constructed firstly. D. H. Gates, et al. suggested that biomechanical state spaces constructed using positions and velocities, or delay reconstruction of individual states, were likely to provide more consistent results [16]. In this study we used each of the three rotation angles time series of lower extremity joints and its time-delayed copies to construct state space,

$$X(t)=[x(t), x(t+\tau), \dots, x(t+(m-1)\tau)], \quad (1)$$

where x(t) was the original data, X(t) was m-dimensional state vector, τ was time delay, and m was embedding dimension. Data set was collected as 30 consecutive strides by 100Hz and re-sampled to 3000 samples suggested by previous studies [17].

Time delays were calculated as the first minimum of the average mutual information function of each signal [12]. In this study, time delay was defined as τ=10. M. B. Kennel, et al. suggested embedding dimension m=5 for all three rotational angles in global false neighbors analysis [18]. D. H. Gates, et al. expected that a minimum of 3 to 6 states would reconstruct the state space with minimal error [16]. In this study, embedding dimension m was defined as 5.

The maximum Lyapunov exponent, that is, λ for human passive walking system could be determined from

$$d_j(i)=d_j(0)e^{\lambda t}, \quad (2)$$

where $d_j(i)$ was the mean Euclidean distance between neighboring trajectories in the reconstructed state space X(t). And it was calculated for every data point j in state space after i discrete time steps. $d_j(0)$ was the initial separation between neighboring trajectories for data point j in state space.

Then, λ was estimated from best fit linear slopes of these local divergence curves defined,

$$y(i)=\frac{1}{\Delta t}\langle \ln[d_j(i)] \rangle, \quad (3)$$

and over scaling regions of time between 0 and 1 stride, that is λ_S. $\langle \ln[d_j(i)] \rangle$ was the average over all values of j.

D. Statistical analysis

As the amputees' amputation was not all on the same side, it is reasonable to classify the amputees' limbs into the sound one and the prosthetic one to compare with the controls' left one and right one. One-way ANOVA test was performed on λ_S derived from three directions of each lower joint angle between the amputees and the controls during three walking conditions. Thus, the difference of walking stability between the amputees' and the non-amputees' walking under equal condition could be achieved. In order to gain the variation between amputees slope walking stability and their own level walking, paired-t test was performed on λ_S derived from three directions of each lower joint angle between the amputees' slope walking performance and their own level walking performance.

III. RESULT

A. Amputees' Walking vs. Controls' Walking

During uphill walking, amputee group showed significantly lower λ_S compared with control group in hip F/E ($P=0.04$), hip IR/ER ($P=0.01$) and ankle F/E ($P<0.01$) in the sound limb side. The amputees also showed significantly lower λ_S in hip F/E ($P=0.01$) compared with the controls in the prosthetic limb side. No significant difference of λ_S was found in any direction of hip motion in the sound limb, knee and ankle motion in the prosthetic limb between two groups. (Table Ⅰ)

During downhill walking, compared with the controls, the amputees showed significantly lower λ_S in knee F/E ($P=0.02$) and ankle F/E ($P=0.03$) in the sound limb, and hip F/E ($P=0.03$) in the prosthetic limb. No significant difference of λ_S

was found in any direction of hip motion on the sound limb, knee and ankle motion on the prosthetic limb side. (Table Ⅱ)

TABLE I. COMPARATION ON Λ_S OF LOWER JOINTS DURING UPHILL WALKING BETWEEN CONTROLS AND AMPUTEES

	Mean λ_S (S.D.)	*Control*	*Transtibial*	*P-value*
Sound limb				
Hip	F/E	2.62(0.36)	2.19(0.35)	0.04*
	Abd/Add	2.20(0.27)	1.95(0.31)	0.14
	IR/ER	1.91(0.32)	1.46(0.19)	0.01*
Knee	F/E	2.31(0.25)	2.03(0.32)	0.10
	Abd/Add	1.96(0.49)	1.91(0.37)	0.83
	IR/ER	1.86(0.44)	1.77(0.27)	0.67
Ankle	F/E	2.14(0.24)	1.67(0.26)	0.00*
	Abd/Add	1.87(0.19)	1.62(0.25)	0.06
	IR/ER	1.87(0.19)	1.62(0.24)	0.05
Prosthetic limb				
Hip	F/E	2.57(0.19)	2.16(0.27)	0.01*
	Abd/Add	2.14(0.31)	1.88(0.21)	0.10
	IR/ER	1.67(0.21)	1.72(0.26)	0.69
Knee	F/E	2.29(0.33)	2.12(0.30)	0.34
	Abd/Add	2.03(0.33)	2.01(0.28)	0.88
	IR/ER	1.77(0.28)	1.64(0.40)	0.52
Ankle	F/E	2.05(0.20)	2.02(0.42)	0.87
	Abd/Add	1.83(0.21)	1.64(0.29)	0.19
	IR/ER	1.83(0.20)	1.66(0.30)	0.23

F/E: flexion/extension; Abd/Add: abduction/adduction; IR/ER: internal rotation/external rotation; Symbol* means significantly different

TABLE II. COMPARATION ON Λ_S OF LOWER JOINTS DURING DOWNHILL WALKING BETWEEN CONTROLS AND AMPUTEES

	Mean λ_S (S.D.)	*Control*	*Transtibial*	*P-value*
Sound limb				
Hip	F/E	2.36(0.26)	2.13(0.34)	0.18
	Abd/Add	2.11(0.26)	1.87(0.28)	0.12
	IR/ER	1.72(0.43)	1.53(0.22)	0.33
Knee	F/E	2.42(0.34)	1.96(0.28)	0.02*
	Abd/Add	1.79(0.36)	1.88(0.35)	0.65
	IR/ER	1.75(0.27)	1.66(0.29)	0.56
Ankle	F/E	1.89(0.31)	1.56(0.15)	0.03*
	Abd/Add	1.92(0.31)	1.65(0.32)	0.14
	IR/ER	1.93(0.32)	1.66(0.33)	0.14
Prosthetic limb				
Hip	F/E	2.42(0.28)	2.09(0.22)	0.03*
	Abd/Add	2.08(0.28)	1.81(0.26)	0.08
	IR/ER	1.58(0.25)	1.57(0.29)	0.92
Knee	F/E	2.28(0.35)	2.08(0.30)	0.27
	Abd/Add	1.97(0.35)	2.00(0.28)	0.87
	IR/ER	1.64(0.29)	1.62(0.43)	0.92
Ankle	F/E	1.86(0.35)	2.00(0.33)	0.48
	Abd/Add	1.82(0.23)	1.61(0.32)	0.18
	IR/ER	1.82(0.25)	1.61(0.31)	0.20

F/E: flexion/extension; Abd/Add: abduction/adduction; IR/ER: internal rotation/external rotation; Symbol* means significantly different

B. Amputees' Slope Walking vs. Level Walking:

While comparing the amputees' uphill walking with their own level walking performance, significantly lower λ_S was found in ankle F/E of the sound limb side (P=0.01). No significant difference of λ_S was found in any direction of the lower joints motion in the prosthetic limb, neither did hip nor knee joint in the sound limb. (Table Ⅲ)

As for the amputees' downhill walking, it showed significantly lower λ_S in ankle F/E in the sound limb (P=0.02) while comparing with their own level walking. No significant difference of λ_S was found in any direction of the lower joints motion in the prosthetic limb, the same as hip and knee joint in the sound limb. (Table Ⅳ)

TABLE III. COMPARATION ON Λ_S OF LOWER JOINTS OF AMPUTEES DURING LEVEL WALKING VS. UPHILL WALKING

	Mean λ_S (S.D.)	*Level*	*Uphill*	*P-value*
Sound limb				
Hip	F/E	2.22(0.35)	2.19(0.35)	0.82
	Abd/Add	2.08(0.44)	1.95(0.31)	0.36
	IR/ER	1.76(0.44)	1.46(0.19)	0.10
Knee	F/E	2.24(0.34)	2.03(0.32)	0.17
	Abd/Add	2.10(0.60)	1.91(0.37)	0.33
	IR/ER	1.72(0.24)	1.77(0.27)	0.23
Ankle	F/E	2.14(0.52)	1.67(0.26)	0.01*
	Abd/Add	1.64(0.35)	1.62(0.25)	0.87
	IR/ER	1.64(0.34)	1.62(0.24)	0.77
Prosthetic limb				
Hip	F/E	2.24(0.36)	2.16(0.27)	0.48
	Abd/Add	2.14(0.47)	1.88(0.21)	0.10
	IR/ER	1.78(0.43)	1.72(0.26)	0.63
Knee	F/E	2.11(0.33)	2.12(0.30)	0.92
	Abd/Add	2.05(0.53)	2.01(0.28)	0.76
	IR/ER	1.79(0.28)	1.64(0.40)	0.47
Ankle	F/E	1.98(0.48)	2.02(0.42)	0.75
	Abd/Add	1.69(0.37)	1.64(0.29)	0.74
	IR/ER	1.68(0.36)	1.66(0.30)	0.85

F/E: flexion/extension; Abd/Add: abduction/adduction; IR/ER: internal rotation/external rotation; Symbol* means significantly different

TABLE IV. COMPARATION ON Λ_S OF LOWER JOINTS OF AMPUTEES DURING LEVEL WALKING VS. DOWNHILL WALKING

	Mean λ_S (S.D.)	*Level*	*Downhill*	*P-value*
Sound limb				
Hip	F/E	2.22(0.35)	2.13(0.34)	0.57
	Abd/Add	2.08(0.44)	1.87(0.28)	0.26
	IR/ER	1.76(0.44)	1.53(0.22)	0.16
Knee	F/E	2.24(0.34)	1.96(0.28)	0.08
	Abd/Add	2.10(0.60)	1.88(0.35)	0.16
	IR/ER	1.72(0.24)	1.66(0.29)	0.52
Ankle	F/E	2.14(0.52)	1.56(0.15)	0.02*
	Abd/Add	1.64(0.35)	1.65(0.32)	0.90
	IR/ER	1.64(0.34)	1.66(0.33)	0.89
Prosthetic limb				
Hip	F/E	2.24(0.36)	2.09(0.22)	0.29
	Abd/Add	2.14(0.47)	1.81(0.26)	0.10
	IR/ER	1.78(0.43)	1.57(0.29)	0.18
Knee	F/E	2.11(0.33)	2.08(0.30)	0.82
	Abd/Add	2.05(0.53)	2.00(0.28)	0.79
	IR/ER	1.79(0.28)	1.62(0.43)	0.46
Ankle	F/E	1.98(0.48)	2.00(0.33)	0.88
	Abd/Add	1.69(0.37)	1.61(0.32)	0.59
	IR/ER	1.68(0.36)	1.61(0.31)	0.64

F/E: flexion/extension; Abd/Add: abduction/adduction; IR/ER: internal rotation/external rotation; Symbol* means significantly different

IV. DISCUSSION

The present study showed that the Lyapunov exponents λ_S were generally lower for the amputee group during slope walking, which means the amputees' lower joints activities were more stable. This result was contrary to our common sense that lower limb amputees would walk more unstably. However, lower limb amputees would adjust their lower joint angles as a compensation strategy during slope walking. Previous researches reported that amputees demonstrated reduced speed, motion range of knee and hip, and hip moments along with increased amplitude and periods of muscle activation during slope walking compared with the controls [19]. We presumed that the amputees confronted much more motion control challenge in lower joints during slope walking that can lead to their decrease in range of joints activity as compensation. Thus, the local dynamic stability of amputees' lower joints during slope walking was significantly improved. This presumption was proved in the comparison of amputees' level walking and slope walking in the present study that better local dynamic stability was got during slope walking. Volunteers involved in this study had rehabilitated for at least one year. Therefore, This result correlates well with previous finding that after at least one year's rehabilitation, most amputees could gain basic walking ability [20].

For the prosthetic limb, results of this work turned out that the amputees behaved significantly more stably in hip F/E compared with the controls during both uphill and downhill walking. Previous research pointed out that shorter step length in slope walking would reduce hip extension in prosthetic limb which decreased the height difference that the prosthetic limb had to adjust to [7]. Thus, it was reasonable that amputees preformed more stably in hip of the prosthetic limb during slope walking.

For the sound limb, compared with the controls, the amputees in this study behaved more stably in hip F/E, IR/ER during uphill walking, and also more stably in knee F/E and ankle F/E during downhill walking. We hypothesized that these results were caused by reduction of hip moments, and motion range of knee and hip. It also indicated that compared with non-amputees, amputees may encounter more motion control burden in hip joint during uphill walking, and in knee joint during downhill walking. For ankle F/E in amputees' sound limb, better stability was found in comparison with the performance of controls during slope walking. And better stability was also found while comparing amputees' slope walking performance with their own level walking performance. Reduced ankle range of motion and power during slope walking were found in former study. These combined reductions were thought to contribute to inadequate step length and difficulty raising the centre of mass up the incline [19]. We hypothesized that ankle in sound limb always confronted more serious motion control challenge for lower limb amputees so that they reduced motion in ankle and performed more stably then.

We also hypothesized slower walking speed was another cause for the result that amputees performed generally more stably than the controls in the present study. It was published that slower walking speed would result in lower Lyapunov exponent [17]. In this study, amputees generally walked slower than controls. We instructed subjects to walk with their most comfortable speed to get their stability which was most similar to their daily performance. However, it still was a limitation that walking speed was not ideally controlled. We are looking forward to finding out a more suitable experiment design to control walking speed factor in further study.

In conclusion, amputees behaved significantly more stably in hip and ankle joint of sound limb and in hip joint of prosthetic limb during uphill walking. And in downhill walking, amputees also behaved significantly more stably in knee and ankle joint of sound limb and in hip joint of prosthetic limb. These outcomes are related to amputees' motion control challenges and compensation strategy. The results can be instructive for prosthetic design and lower limb amputees' rehabilitation to release amputees' motion control burden in these joints during walking.

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