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Kinematic and Dynamic Analysis of the Gait Cycle of Above-Knee Amputees

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The change of gait pattern and muscular activity following amputation is thought to be responsible for the higher incidence of joint degenerative disorders observed in amputees. Considering the lack of consistent data in the literature, the purpose of the present study was to measure and analyze the spatio-temporal variables, the kinematics and, particularly, the net joint moments of the intact and prosthetic limbs of above knee amputee subjects during walking and to compare the results with those of normals. The gait characteristics of five transfemoral amputees and five normal subjects were measured using videography and a force platform. The human body was modeled as a 2-D sagittal plane linkage consisting of 8 rigid segments and analyzed using rigid body kinematics and inverse dynamics approaches. The results, including the joints flexion angles and normalized net moments, were statistically analyzed. There was a significant difference between the spatio-temporal variables of the normal subjects and intact and prosthetic limbs of amputee subjects, but, the difference between the intact and prosthetic limbs of amputees was not statistically significant. The kinematics of the intact limb of the amputee subjects was close to that of the normals, but the prosthetic limb had a much more limited angular motion. The intact limbs of the amputee subjects experienced larger than normal extension hip moment (with a maximum value of 2.08 compared to 1.68 Nm/kg) and flexion knee moment (with a maximum value of 1.84 compared to 1.14 Nm/kg); this is believed to contribute to the articular cartilage lesions. The hip joint of the prosthetic limb of the amputee subjects, on the other hand, experienced lower than normal joint moments (with a maximum value of 0.97 compared to 1.67 Nm/kg), which might contribute to the osteoporosis found in the remainder of the femur.

INTRODUCTION

The human gait includes a harmonized motion of the lower extremities (feet, legs and thighs) via the coordinated function of the muscles crossing the related joints (ankles, knees and hips). In normal gait, not only is the energy consumption optimized, but also, the resulting loads are regulated so that they can be well tolerated by the joints without initiating destructive changes in the articulating cartilage [1]. Following amputation, however, the normal gait pattern is altered, due to the inevitable structural and functional changes in the musculoskeletal system.

The subsequent effect on the load of the lower extremities is a matter of concern, since it might contribute to the disorders in bones (e.g., osteoporosis) or articulating joints (e.g., osteoarthritis) of both the intact and prosthetic limbs. Clinical studies of Bruke et al. [2], Hungerford and Cockin [3] and Norvell et al. [4] reported a higher incidence of osteoarthritis at the knee joint of the intact limbs of transtibial and transfemoral amputee subjects. Recent studies of Kulkani et al. [5] and Rush et al [6], on the other hand, have shown that the occurrence rate of osteopenia/osteoporosis is significantly higher on the amputee side than in the intact sides of the above knee amputees.

The difference in joint loads of the intact and amputee limbs of below knee amputees, in comparison with those of normal subjects, has been extensively studied in the literature. Lewallen et al. [7] studied

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the net joint moments of the knee and hip of a group of below knee amputee children (4 to 13 year old). They reported that the knee and hip joint moments of the intact limbs were normal or below normal during walking and attributed this to a slower walking speed, decreased step length, decreased double support time and increased stance time. Hurley et al. [8] studied the net knee and ankle joints reaction forces of a group of below knee amputee subjects (24 to 46 years of age). They reported no significant difference for vertical forces but a significantly lower than normal peak horizontal force for amputee subjects, due to a slower walking speed and a lower push off force at the prosthetic side. Lemaire and Fisher [9] studied the net joint moments of an elderly group of transtibial amputee subjects and a similar control group of normals. They reported that the amputee group experienced consistently larger knee joint net moments at their intact limbs.

However, in contrast to the below knee amputation literature, relatively few reports of the kinematics and kinetics of above knee amputee gaits have been published. Moreover, most of these studies [10-14] have been limited to the evaluation of the temporal parameters and kinematics of above knee amputees rather than dynamics and joint loads.

The few previous studies addressing the joint loads of above knee amputees are often based on the gait analysis of only few subjects and have revealed no results for the intact limbs and/or, statistically, inconsistent results for the prosthetic limbs. Yang et al. [15] studied the influence of alignment of the prosthesis on the gait of a total of 4 above knee amputees and reported the joint moment results for only one subject. Seroussi et al. [16] compared the gait parameters of eight normal and eight above knee amputee subjects, wearing the same lightweight prostheses. Although not tested statistically, they reported a higher than normal hip and a lower than normal knee joint moment for the intact limb, and a higher than normal hip flexor moment for the prosthetic limb of amputee subjects. Blumentritt et al. [17] evaluated the kinematics and dynamics of seven transfemoral amputees wearing an Otto Bock 3R80 prosthetic knee. They reported that the knee and hip joint moments were highly variable among different subjects. The peak knee extension moment ranged from 20 to 53 Nm on the prosthetic side and 41 to 58 Nm on the intact side. The hip maximum extension moment during the stance phase varied between 25 and 105 Nm and the maximum pre-swing hip flexion moment ranged from 27 to 59 Nm. Van der Linden et al. [18] analyzed the gait pattern of two male transfemoral amputees and reported highly variable joint moments for both prosthetic and intact limbs when wearing four different prosthetic feet. Stephenson and Seedhom [19] reported the force and moments at the interface between an artificial limb and an implant directly fixed into the femur of transfemoral amputees. Perry et al. [20] studied the energy expenditure and gait characteristics of a single bilateral amputee wearing C-leg prostheses and reported that the hip and knee joint moments were two thirds and one third of the normals, respectively. Yokoqushi et al. [21] compared the biomechanical gait parameters of three transfemoral amputees wearing, alternatively, a newly designed polycentric knee and a polycentric hydraulic knee prosthesis with those of 10 healthy volunteers and reported a higher peak for the hip extension moment of the amputees in comparison with the normals.

Considering the lack of consistent data for the joint loads of prosthetic and intact limbs of above knee amputees, the purpose of the present study is to quantitatively analyze and compare the dynamic characteristics of the gait pattern of above knee amputee and normal subjects, including spatiotemporal variables, kinematics and, particularly, net joint moments of ankle, knee and hip joints, using the inverse dynamic method.

METHODOLOGY

The tests were conducted on two groups of subjects. The amputee subjects included four men with transfemoral amputation and one man with knee disarticulation. They had an average age of 37.8 ± 4.48 years, height of 1.74 ± 0.09 m, mass of 72.4 ± 11.97 kg and Body Mass Index (BMI) of 24.63 ± 2.99 kg/m^2 (Table 1). Before beginning the test, all subjects were examined by a prosthetist to ensure that all prostheses were well aligned and comfortable for the subjects and none of the subjects had residual limb problems, e.g., pain, swelling, pressure sores, painful motion, crepitus with motion, ligamentous instability, limitation of motion, etc. Detailed characteristics of the prosthetic limbs used by the amputee subjects is shown in Table 2.

The normal subjects included five men with similar BMI and without any lower limb pathology. Their average age, height, mass and BMI were 24.2 ± 0.83 years, 1.74 ± 0.06 m, 74.4 ± 6.65 kg and 24.55 ± 2.99 kg/m^2, respectively. None of the subjects had a history of orthopedic, neurological, cardiovascular and/or respiratory problems, except for traumatic amputation of the amputee group. Normal and amputee groups were matched by BMI for statistical comparison.

The body of each subject was modeled as a two dimensional sagittal plane linkage, consisting of 8 rigid segments, i.e., trunk, pelvis, and left and right feet, legs and thighs, connected at hinge joints. During the tests, the subjects wore tight fitting black clothes with reflective markers (0.02 m diameter) fixed onto

Table 1. Amputee subjects' data.

Subjects / Variable	1	2	3	4	5
Age (yr)	41	34	44	35	35
Height (m)	1.70	1.65	1.7	1.83	1.86
Body Weight (kg)	62	79	58	85	82
Body Mass Index (kg/m^2)	21.45	29.01	20.07	25.38	24.57
Duration of Amputation (yr)	19	16	20	17	16
RL Length (cm)	19	25	18	19	Knee disarticulation
Side	Left	Left	Right	Left	Left
Use Cane	No	No	No	No	No

Table 2. Prostheses data.

Variable / Subjects	Socket	Suspension	Knee	Foot
1	Ischial weight bearing	Suction & silesian strap	3R20	Single axis
2	Ischial weight bearing	Suction	3R15	Single axis
3	Ischial weight bearing	Suction & silesian strap	3R15	Single axis
4	Ischial weight bearing	Suction & silesian strap	3R20	Single axis
5	Ischial weight bearing	Suction	3R21	Single axis

anatomical landmarks at the left and right acromion process (shoulder joint), anterior superior iliac spine, greater trochanter (hip joint), lateral femoral epicondyle (knee joint), lateral malleolus (ankle joint), the 2nd metatarsal head and L5-S1 joint (along anterior superior iliac spine). For the prosthetic limb of the amputee subjects, joint centers of the foot and knee were estimated, based on the corresponding locations at the sound limb [22,23].

While the arm swing was controlled, each subject walked at a free cadence along a 6.5 m walkway, with a force plate at the midpoint and a black screen in the background. The screen had 3 points at the corners of an isosceles right triangle as the 2D reference coordinate system. The walking cycle was filmed at a 25 fps speed, using a digital camera (Sony DCR-TRV 330 E EIS), mounted perpendicular and 10 m far from the walkway at knee height. The motion data was then transferred to a PC through a digital video I link (IEEE 1394 standard) fire wire and combined with the force data obtained by the force plate (Kistler 9286A) at 50 Hz. The camera and force plate data were synchronized using a LED which flashed as the force plate started data collection.

On average, 35-45 trials were recorded for each subject; the subjects were asked to stop the test for a 5 minute rest whenever they felt tired. Among the recorded trials, the first 10 and the ones with an incomplete contact between the camera facing foot and force plate were omitted. Then, two trials for each limb of the amputee subjects and two trials for one side of the normal subjects were selected randomly to be analyzed. The markers were traced using Sharif Motion Analysis software, obtaining their two-dimensional coordinates (Figure 1). All body segment anthropometric parameters were derived from the regression relationships established by Dempster

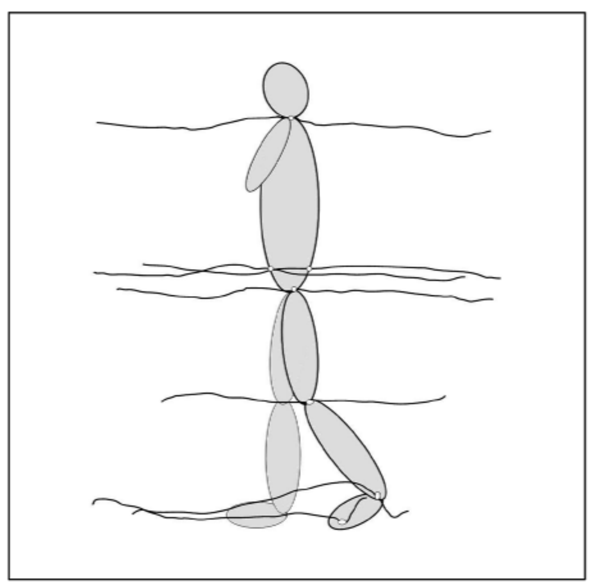

Figure 1. The markers trajectory during the gait cycle.

and reported by Winter [24]. The parameters of the prosthetics side were assumed to be the same as the intact side [23].

First, for accessing external joint moments, linear and angular velocities and accelerations were calculated using the finite difference method and rigid body kinematics equations. The markers linear velocity and acceleration were obtained using Equations 1 and 2 (except for the first and last frames, for which the data of the two first and two last frames were used, respectively). Each segment's angular velocity and acceleration were calculated using Equations 3 and 4 and its center of mass linear acceleration components using Equations 5 and 6:

$$\dot{x}_n = \frac{x_{(n+1)} \quad x_{(n\quad 1)}}{t_{(n+1)} \quad t_{(n\quad 1)}},$$

$$\dot{y}_n = \frac{y_{(n+1)} \quad y_{(n\quad 1)}}{t_{(n+1)} \quad t_{(n\quad 1)}}, \tag{1}$$

$$\ddot{x}_n = \frac{\dot{x}_{(n+1)} \quad \dot{x}_{(n\quad 1)}}{t_{(n+1)} \quad t_{(n\quad 1)}},$$

$$\ddot{y}_n = \frac{\dot{y}_{(n+1)} \quad \dot{y}_{(n\quad 1)}}{t_{(n+1)} \quad t_{(n\quad 1)}}, \tag{2}$$

$$\omega_n = \frac{(\dot{x}_{Dn} \quad \dot{x}_{Pn})}{y_{Dn} \quad y_{Pn}}, \tag{3}$$

$$\alpha_n = \frac{[\ddot{x}_{Dn} \quad \ddot{x}_{Pn} + \omega_n^2.(x_{Dn} \quad x_{Pn})]}{y_{Dn} \quad y_{Pn}}, \tag{4}$$

$$\ddot{x}_{Cn} = \ddot{x}_{Pn} \quad \omega_n^2.(x_{Cn} \quad x_{Pn}) \quad \alpha_n.(y_{Cn} \quad y_{Pn}), \tag{5}$$

$$\ddot{y}_{Cn} = \ddot{y}_{Pn} \quad \omega_n^2.(y_{Cn} \quad y_{Pn}) \quad \alpha_n.(x_{Cn} \quad x_{Pn}). \tag{6}$$

To obtain the net joint forces and moments, the inverse dynamics method was employed. At first, the joint forces and moments in the most distal joint (ankle) were calculated using the dynamic equilibrium equations of the foot segment (Figure 2):

$$F_{x\text{Ankle}} = (m_{\text{Foot}}.\ddot{x}_{C.\text{Foot}}) \quad R_x, \tag{7}$$

$$F_{y\text{Ankle}} = (m_{\text{Foot}}.\ddot{y}_{C.\text{Foot}}) + (m_{\text{Foot}}.g) \quad R_y, \tag{8}$$

$$\begin{aligned} M_{\text{Ankle}} = {} & (F_x.y_{c.p}) \quad (F_y.x_{c.p}) \\ & (m_{\text{Foot}}.\ddot{x}_{C.\text{Foot}}.y_{C.\text{Foot}}) \\ & + (m_{\text{Foot}}.\ddot{y}_{C.\text{Foot}}.x_{C.\text{Foot}}) \\ & + (m_{\text{Foot}}.g.x_{C.\text{Foot}}) + (F_{x\text{Ankle}}.y_{\text{Ankle}}) \\ & (F_{y\text{Ankle}}.x_{\text{Ankle}}) + (I_{\text{Foot}}.\alpha_{\text{Foot}}). \end{aligned} \tag{9}$$

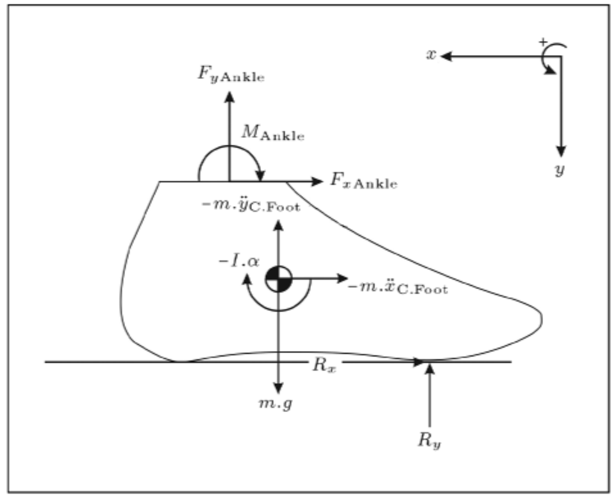

Figure 2. Free body diagram of the right foot segment.

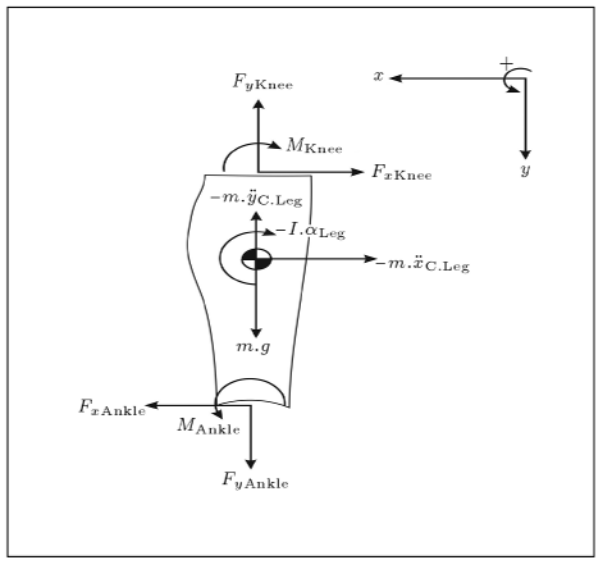

Figure 3. Free body diagram of the right shank segment.

Then, considering Figure 3, the knee joint net force and moment were obtained using dynamic equilibrium equations for the shank segment:

$$F_{x\text{Knee}} = (m_{\text{Leg}}.\ddot{x}_{C.\text{Leg}}) + F_{x\text{Ankle}}, \tag{10}$$

$$F_{y\text{Knee}} = (m_{\text{Leg}}.\ddot{y}_{C.\text{Leg}}) + (m_{\text{Leg}}.g) + F_{y\text{Ankle}}, \tag{11}$$

$$\begin{aligned} M_{\text{Knee}} = {} & \quad M_{\text{Ankle}} \quad (F_{x\text{Ankle}}.y_{\text{Ankle}}) \\ & + (F_{y\text{Ankle}}.x_{\text{Ankle}}) + (F_{x\text{Knee}}.y_{\text{Knee}}) \\ & (F_{y\text{Knee}}.x_{\text{Knee}}) \quad (m_{\text{Leg}}.\ddot{x}_{C.\text{Leg}}.y_{C.\text{Leg}}) \\ & + (m_{\text{Leg}}.\ddot{y}_{C.\text{Leg}}.x_{C.\text{Leg}}) + (m_{\text{Leg}}.g.x_{C.\text{Leg}}) \\ & + (I_{\text{Knee}}.\alpha_{\text{Leg}}). \end{aligned} \tag{12}$$

And, finally, for the hip joint, the net joint moment was calculated using dynamic equilibrium equations of the thigh segment (Figure 4):

$$\begin{aligned} M_{\text{Hip}} = \quad & M_{\text{Knee}} \quad (F_{x\text{Knee}}.y_{\text{Knee}}) + (F_{y\text{Knee}}.x_{\text{Knee}}) \\ & + (F_{x\text{Hip}}.y_{\text{Hip}}) \quad (F_{y\text{Hip}}.x_{\text{Hip}}) \\ & \quad (m_{\text{Thigh}}.\ddot{x}_{C.\text{Thigh}}.y_{C.\text{Thigh}}) \\ & + (m_{\text{Thigh}}.\ddot{y}_{C.\text{Thigh}}.x_{C.\text{Thigh}}) \\ & + (m_{\text{Thigh}}.g.x_{C.\text{Thigh}}) + (I_{\text{Hip}}.\alpha_{\text{Thigh}}). \quad (13) \end{aligned}$$

All the moment equilibrium Equations 9, 12 and 13 were written about the origin of the coordinate system.

To be able to compare the results of different subjects, the joint moments of each subject were normalized by their body weight to eliminate the effect of weight differences. Also, the results related to each time frame were normalized as a percent of the subject's whole stride duration, to eliminate the effect of walking speed differences. Finally, to be able to interpolate and average the results of a group and compare between different groups, a cubic spline curve was fitted to the relevant data using MATLAB.

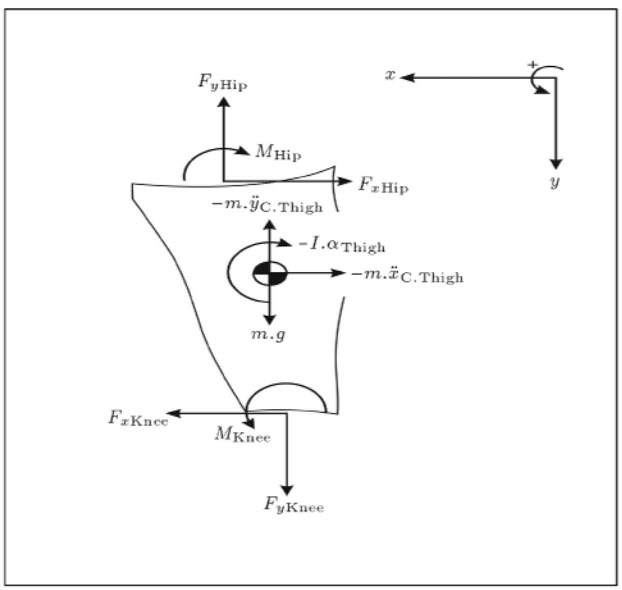

Figure 4. Free body diagram of the right thigh segment.

Statistical comparison between means of spatio-temporal and joint moment variables of two limbs of amputee subjects was conducted using the paired t-test (parametric method) and the Wilcoxon Signed Ranks Test (nonparametric method). Statistical comparison between the variables of normal subjects and each limb of the amputee subjects was conducted separately using the t-test (parametric method) and the Mann-Whitney test (nonparametric method). These tests were selected, based on the recommendations of a statistics advisor and the literature [9,10,12,21].

RESULTS

The gait spatio-temporal variables for the normal subjects and intact and prosthetic limbs of amputee subjects are shown in Table 3. The stride length, step length and stepping speed (mean velocity of the foot during a step) were significantly higher and the step duration (time duration of a step) was significantly lower for normal subjects in comparison with the prosthetic and sound limbs of the amputee subjects ($p < 0.05$). These variables, however, were not significantly different for the prosthetic and sound limbs of amputee subjects, except for the stance duration.

The knee and hip joint angles were considered as the flexion angle of the relevant joints; the angle between shank-thigh and thigh-pelvis segments in the sagittal plane, respectively, while their magnitudes assumed zero at a standing position (anatomical posture) and negative in extension. The ankle angle was considered as the dorisflexion angle of the ankle joint; the angle between foot-shank segments in the sagittal plane, with a zero magnitude in anatomical posture and negative magnitude in plantarflexion.

The variation of the ankle joint angle for the normal subjects and prosthetic and intact limbs of amputee subjects is shown in Figure 5a. The general pattern of the ankle joint angle is similar within the

Table 3. Comparison of means of spatio-temporal variables.

Subjects	Normal Subjects		Amputee Subjects			
			Intact Limb		Prosthetic Limb	
Variable	Mean	SD	Mean	SD	Mean	SD
Stride Length (m)	1.238	0.07	1.003	0.12	1.003	0.12
Step Length (m)	0.62	0.034	0.46	0.08	0.49	0.11
Stance Duration (%)	59.47	2.938	65	2.26	60.36	3.12
Stepping Speed (m/s)	0.97	0.07	0.68	0.228	0.7	0.15
Walking Speed	0.96	0.068	0.668	0.196	0.668	0.196

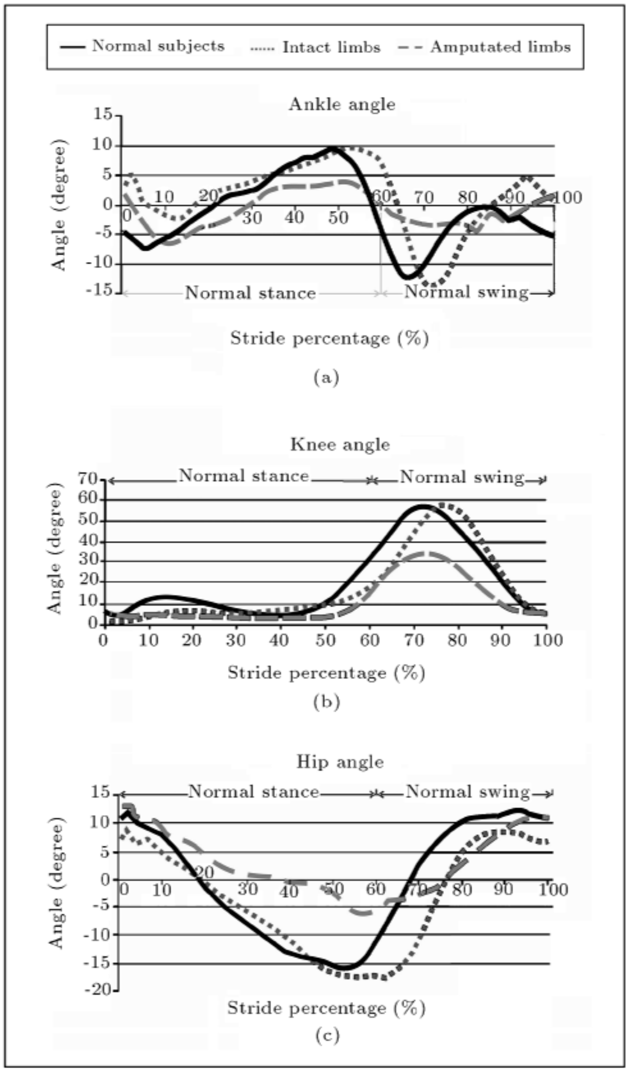

Figure 5. Variation patterns of the mean ankle (a), knee (b) and hip (c) joint angle for normal subjects and intact and prosthetic limbs of amputee subjects.

normal subjects (±1 5° St Deviation), and the prosthetic limbs of amputee subjects (±0.5 5° St Deviation), but different among the intact limbs of amputee subjects (±1 11° St Deviation), especially during the swing phase. On the mean curve of the normal subjects, there was a small plantarflexion after the initial contact, with a peak value of -7° at 6% of the stride, which then transformed to dorsiflexion, with a maximum of +9° at 49% of the stride. During the swing phase, the ankle went to plantarflexion again, with a peak of -12° at 66% of the gait cycle.

For the intact limbs of the amputee subjects, the variation of the ankle joint angle was similar to that of the normal subject, but with a short time delay. The initial plantarflexion was smaller, with a peak of -2° occurring at 13% of the stride and the following dorsiflexion and plantarflexion were larger, with maximum values of +9° and -14° occurring at 53% and 72% of the gait cycle, respectively. For prosthetic limbs of amputee subjects, the general pattern of the mean curve was similar to those of the intact limbs and normal subjects, but with small deviations from the neutral position and quite low peak values. This included a -6° plantarflexion at 10% and a +4° dorsiflexion at 50% of the gait cycle.

The knee joint angle graphs for the normal subjects and for the prosthetic and intact limbs of amputee subjects are shown in Figure 5b. The positive and negative directions indicate flexion and extension, respectively. The general pattern of the knee joint angle was similar within the normal subjects (±1° St Deviation in stance phase and ±10° in swing phase), and the intact limbs of the amputee subjects (except for one case), but quite variable among the prosthetic limbs of the amputee subjects, especially during the swing phase.

Two flexion waves were observed on the mean curve of the normal subjects knee angle, including a stance phase flexion peak of +13° at 14% and a swing phase flexion peak of +56° at 72% of the gait cycle. The same pattern was observed for the intact limbs of the amputee subjects with a short time delay. However, the prosthetic limbs remained extended during the stance phase. The first flexion peaks were smaller for the intact limbs than those of the normal subjects, including a +7° flexion at 20% of the stride. The swing phase flexion peak was close to that of the normal subject for the intact limb (+57° at 78% of the gait cycle) but much lower for the prosthetic limb (+34.5° at 73% the gait cycle).

The variation of the hip joint angle for the normal subjects and for the prosthetic and intact limbs of amputee subjects are shown in Figure 5c. The positive and negative directions indicate flexion and extension, respectively. The general pattern of the hip angle is similar within the normal subjects and the intact limbs of the amputee subjects, but, quite variable among the prosthetic limbs of the amputee subjects. On the mean curve of the normal subjects, there was a slight flexion at the initial stance (with a peak of +12°), followed by a large extension during the rest of the stance phase (with a peak of -16° at heel off), which returned to flexion from 80% to the end of the gait cycle (with a peak of +12°).

For the intact limbs of the amputee subjects, the variation of the hip joint angle was similar to that of the normal subject, but, the initial and terminal flexions were smaller (with equal peak values of 9°) and the middle extension was larger (with a peak of 17° during 51% to 64% of the gait cycle). The general pattern of the hip joint angle, however, was different for the prosthetic limbs of amputee subjects. The hip joint remained in flexion during most of the stance phase, with a maximum of 13° at initial contact. Also, the maximum extension was only -6° and the terminal flexion continued to the end of the gait cycle.

The ankle net joint moment graphs for normal subjects and for the prosthetic and intact limbs of the amputee subjects are shown in Figure 6a. The positive and negative directions indicate plantarflexor (extensor) and dorsiflexor (flexor) moments, respectively. The general pattern of the ankle joint moments was similar within normal and prosthetic limbs, but variable among the intact limbs of amputee subjects. On the mean curve of the normal subjects, there was a slight dorsiflexor moment of -0.04 Nm/kg immediately after initial contact, which rapidly reversed to a plantarflexor moment in the remainder of the stance phase. The peak plantarflexor moment occurred at 49% of the stride with a value of +1.67 Nm/kg (±0.16 St Deviation) and, then, gradually fell to zero at 62% of the stride.

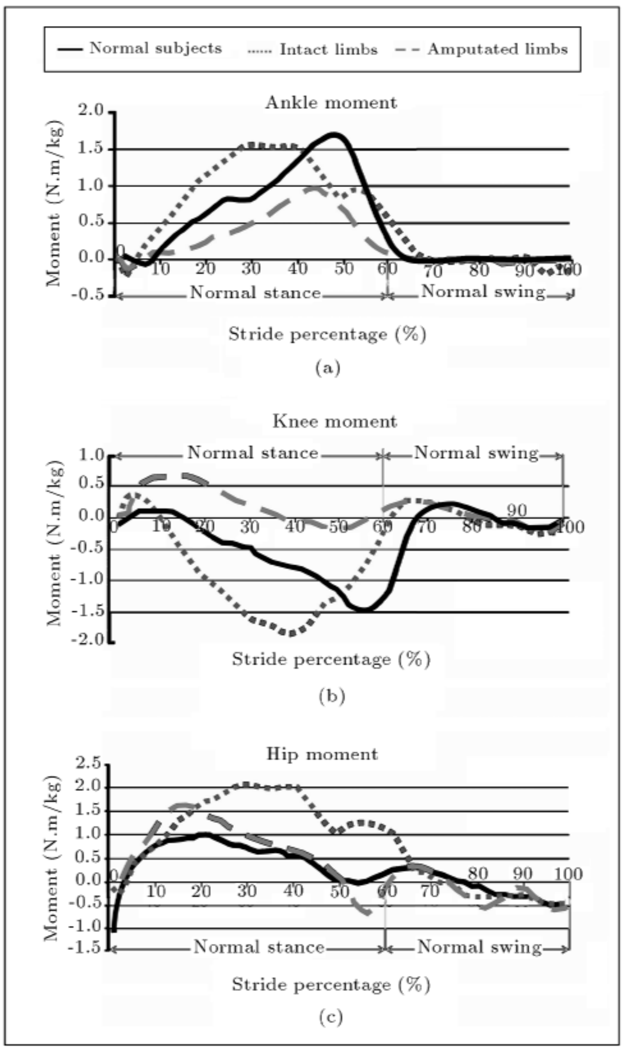

Figure 6. Variation patterns of the mean ankle (a), knee (b) and hip (c) joint net moments for normal subjects and intact and prosthetic limbs of amputee subjects.

For the intact limbs of amputee subjects, the peak of plantarflexor moment and its incidence time were very different among the subjects. This caused a flat region from +1.38 Nm/kg to +1.58 Nm/kg (±1.33 to 1.38 St Deviations) at 30% to 40% of the gait cycle on the mean curve, which then came to zero at 70% of the stride with a gradual decline. For prosthetic limbs, much smaller ankle net joint moments were observed. The peak plantarflexor moment was only +0.96 Nm/kg (±0.54 St Deviation) occurring at 44% of the stride, which gradually fell to zero at 67% of the gait cycle.

The knee net joint moment graphs for normal subjects and for the prosthetic and intact limbs of amputee subjects are shown in Figure 6b. The positive and negative directions indicate extensor and flexor knee moments, respectively. The general pattern of the knee joint moment was nearly similar within each subject group. On the mean curve of normal subjects, there was a small extensor moment after the initial contact, with a peak of +0.13 Nm/kg, occurring at 8% of the stride. This was then transformed to a flexor moment with a maximum of -1.14 Nm/kg (±1.14 St Deviation) occurring at 50% of the stride. The knee joint moment then reversed again to an extensor moment with a peak of +0.25 Nm/kg (±0.07 St Deviation) at 65% of the gait cycle and fell down to approach zero in the remainder of the gait cycle.

For the intact and prosthetic limbs of amputee subjects, the knee initial extensor moments were greater than those of normal subjects, with peak values of +0.32 Nm/kg (±0.29 St Deviation) and +0.69 Nm/kg (±0.44 St Deviation), respectively. The following flexor moment was also very large for intact limbs with a peak of -1.84 Nm/kg (±2.29 St Deviation), but quite small for the prosthetic limb with a peak of -0.2 Nm/kg (±0.36 St Deviation). Both limbs then exhibited extensor moments with peak values close to the normal subjects, which gradually declined to zero.

The hip net joint moment graphs for normal subjects and for the prosthetic and intact limbs of amputee subjects are shown in Figure 6c. The positive and negative directions indicate extensor and flexor hip moments, respectively. The general pattern of the hip joint moments was nearly similar within each subject group. On the mean curve of normal subjects, there was a flexor moment of -0.42 Nm/kg at the initial contact that gradually reversed to an extensor moment with a maximum of +1.67 Nm/kg (±0.96 St Deviation) at 16% of the stride. The hip joint moment then transformed to a flexor moment with a peak of -0.67 Nm/kg (±0.95 St Deviation) occurring at 56% of the gait cycle and, then, again became an extensor moment with a peak of 0.29 Nm/kg (±0.49 St Deviation) occurring at 64% of the stride. The moment then remained a small fluctuating flexor moment until the end of the gait cycle.

For the intact and prosthetic limbs of amputee subjects, the hip joint moment was an extension moment in the major part of the gait cycle. The magnitude of this moment, however, was quite different in the two subject groups. For the intact limbs, it was larger than that of the normal subjects, with a peak of +2.08 Nm/kg (±1.73 St Deviation) occurring at 30% of the stride. For prosthetic limbs, on the other hand, it was much smaller, including a peak of +0.97 Nm/kg (±0.58 St Deviation) at 22% of the gait cycle. Both limbs then exhibited a gradual decline during the remainder of the gait cycle.

DISCUSSION

This study suffers from a number of limitations, e.g., the small number of subjects, the 2-D analysis, the low speed of cinematography, etc. Also, there were some sources of error which might affect the results, e.g., the synchronization method and probable motion of the markers on the clothing of the subjects. It has been shown by Zahedi et al. [25] that such methodological errors, as well as the step to step variation of the subjects' gait, can significantly affect the measured parameters. This is most substantial around the ankle joint, where the degree of kinematic variation for the step to step gait of an individual and among different subjects is higher and, at the same time, a small error in marker tracking can cause a large error in angle measurement, due to the short distance between markers attached to the feet. The results of the present study are obviously affected by such factors. For instance, the plantar/dorsi flexion of the prosthetic ankle during the swing phase (Figure 5a) is due to measurement errors. However, in general, the repeatability tests revealed a good correlation between the data gathered during different trials for each individual subject. The ground reaction force and knee angle variation of one normal subject are shown in Figures 7a and 7b, respectively. Moreover, the general pattern of the results of the present study, for both normal and amputee subjects, corresponds well with those reported by previous studies. For instance, in comparison with the results of the classic works of Winter [26] and Whittle [27] on normal gait, a good agreement is found between the means of most comparable quantities. This will be discussed in more detail, specifically for above knee amputees' gait, for each individual variable.

The significant difference of the gait spatio-temporal variables of normal and amputee subjects observed in the present study correlates well with the significant difference of their walking speed and has been reported by other investigators [10,18]. Among these variables, the stance phase duration obtained for the normal subjects and for the prosthetic and intact limbs of the amputee subjects of the present study were close to the corresponding data in the literature [10,18,21]. However, for the step and stride lengths, although significant differences between the prosthetic and sound limbs of amputees and/or amputee and normal subjects, respectively, has been well documented in the literature [10,18], much larger magnitudes for both limbs or groups have been reported. Obviously, this is due to the much lower free speeds of our subjects (0.96 m/s for normals and 0.64 m/s for amputees) in comparison with those reported in the literature (e.g., 1.51 m/s for normals and 1.07 to 1.20 m/s for amputees in the work of [10]). In fact, the results for free speed walking are quite close to those [10] reported for slow walking speed and may indicate a general trend among the Iranian population.

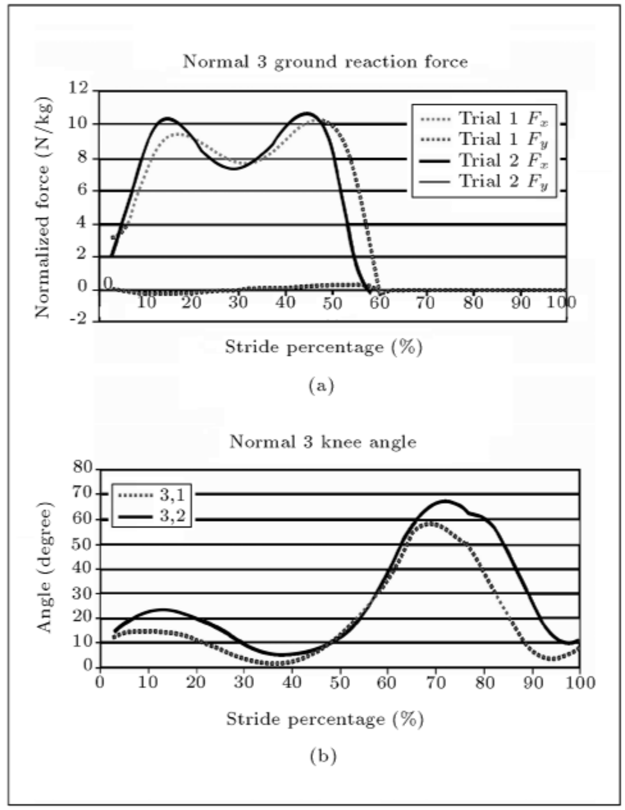

Figure 7. Repeatability of reaction force (a) and kinematical data (b) for one of the normal subjects.

The spatio-temporal variables were not significantly different between the prosthetic and intact limbs of the amputee subjects, except for the stance phase duration. A longer stance phase duration for an intact limb, in comparison with a prosthetic limb, caused an asymmetric walking pattern within the amputee group. It is believed that this helps to compensate the functional and load bearing limitations caused by amputation.

The general patterns of the mean curves of the ankle, knee and hip joint angles that were obtained for normal and amputee subjects (Figure 5) were similar to the results reported in the literature with, however, much smaller ranges of motion. For the ankle joint of the prosthetic limbs (Figure 5a), a smaller than normal plantarflexion peak (after the initial contact) followed by a delayed dorsiflexion peak (near the toe off) has also been reported by Seroussi [16] and Perry [20]. For the knee joint angle of the intact limbs of amputees, a small stance flexion peak (of 7 degrees) and a delayed swing flexion peak was found in comparison with normals (Figure 5b). The results in the literature are not consistent concerning the stance phase flexion peak (10 by Yang [15], but over 20 by Seroussi [16] and Van der Linden [18]). However, a delayed swing flexion peak has been well described in the above references. For the prosthetic limb, the knee joint remained extended during the stance phase; the small variation from 0 is due to errors in the positioning and tracking of markers and has also been reported by Serrussi [16], Perry [20] and Yokogushi [21]. The mean flexion peak obtained for the prosthetic knee during the swing phase was 34.5 (Figure 5b), which is smaller than all previous data ranging between 41 to 70 [10,12,13,17,18,20,21]. Also, the mean hip angle of the amputee subjects varied in a small range of motion (Figure 5c) in comparison with that reported in the literature [12,13,15,20,21]. It is believed that the smaller ranges of motion obtained are due to the slower walking speed of the subjects, in comparison with that of the western subjects analyzed in the literature.

In general, the kinematics of the intact limb of the amputee subjects were close to those of the normal subjects and significantly different from their prosthetic limb. A much more limited angular motion was observed for the prosthetic limb, which is not unexpected considering the lack of muscular function. This might seem in contradiction with the insignificant difference of most spatio-temporal variables observed for these two limbs. The fact is that these variables, e.g., the step and, consequently, stride lengths, are not affected only by the kinematics of a single limb; they are contributed to by the joints of both limbs.

The general pattern of the mean curves of the ankle joint moment obtained in this study (Figure 6a) is similar to the results reported in the literature, particularly for normal subjects and for the prosthetic limbs of amputee subjects [15,16,18,20].

The (mean) peak of plantarflexion moment for the prosthetic limbs was about 60% that of normals and intact limbs; this is in good agreement with the corresponding data in the literature, including 62% [16], 70% [15], 71% [20] and nearly 50% [26]. For the intact limbs of amputee subjects, Seroussi [16] reported a higher plantarflexion peak in comparison with normals; this was not observed in this study.

For the knee joint moment of the intact limbs of amputee subjects, close to normal extensor and 60% higher than normal (mean) flexor peaks were obtained. This is consistent with the results reported by Seroussi [16]. For prosthetic limbs, the moment curve had a fluctuating pattern, including an extensor, then flexor, then extensor moment during an early, middle and late stance, respectively, close to that reported by Perry [20]. A similar pattern has also been reported by Van der Linden [18] for three of the four prosthetic limbs they examined. However, the extensor moment they observed during the late stance and swing phase was much larger than that reported here. Yang [15] and Seroussi [16], on the other hand, have reported a permanent extensor and flexor moment, respectively, during the early and middle stance. A flexor moment at the prosthetic knee joint, during the stance phase, is difficult to justify. Some references have attributed this to the extensor mechanical stop [26] and/or stabilizing mechanism of the stance phase control mechanism of the prosthetic knee [18].

The hip joints of the intact and prosthetic limbs of the amputee subjects experienced extensor moments in the major part of the gait cycle, in opposition to the normal subjects. This is consistent with the results reported by Van der Linden [18]. However, other researchers often reported flexor moments during the late stance [15,16,20]. The extending nature of the hip joint moment of prosthetic limbs has been recognized as a major factor in knee stabilization and propulsive force generation [26]. The results in this paper also suggested 25% higher and 53% lower than normal (mean) peaks for the intact and prosthetic limbs of amputee subjects, respectively. This is in good agreement with the results reported by other investigators [15,16,18,20].

The results of this study suggest that in a comparison between the net joint moments of an intact limb of amputee subjects and of normal subjects, the hip and knee joints of the intact limb of amputees experience consistently larger net moments. Particularly, higher values were found for the maximum joint moments of the hip (with a mean of 25% higher than normal) and knee (with a mean of 60% higher than normal) and the application time of the above-average joint moments. On the other hand, the hip joint of the prosthetic limbs of amputee subjects experienced lower joint moments in comparison with intact limbs (more than 50%) and with normal subjects (more than 40%).

Clinical studies have indicated that mechanical factors, e.g., a higher than normal load on the intact limb, more time spent on the intact limb and a lower than normal load on the amputee limbs, are the main factors responsible for degenerative lesions observed frequently in long-term prosthetics users [2,3,5,6,9,28-30]. This suggests that the risk of articular cartilage damage and osteoarthritis is higher for the intact limbs of amputee subjects in comparison with normal subjects. This conclusion is consistent with the results of the recent retrospective cohort study of Norvell et al. [4] reporting standardized prevalence ratios of 3.3 and 1.3 for pain and symptomatic osteoarthritis, respectively, in the intact knee of amputees, compared with normals. Also, due to a lower load at the hip joint of the prosthetic limb of amputee subjects, the remaining femur of the prosthetic limb is at a higher risk of osteoporosis occurrence.

NOMENCLATURE

C	segment's center of mass
$C.P$	centre of pressure
D	distal joint
F	joint's net force
g	gravity acceleration
I	segment's moment of inertia
M	joint's net moment
m	segment's mass
n	frame number
P	proximal joint
R	ground reaction force
T	time
x	horizontal axis; horizontal component of displacement
$\dot{x}$	horizontal component of linear velocity
$\ddot{x}$	horizontal component of linear acceleration
y	vertical axis; vertical component of displacement
$\dot{y}$	vertical component of linear velocity
$\ddot{y}$	vertical component of linear acceleration
α	angular acceleration
ω	angular velocity

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