



## Original Article

## The Interaction of Knee, Hip and L5-S1 Joint Contact Forces and Spatiotemporal Variables Between Sound and Prosthetic Leg in Patients with Unilateral Below-Knee Amputation During Walking

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### ABSTRACT

**Background:** There is little knowledge that explains how forces are applied on knee, hip and L5-S1 joint between the sound and prosthetic leg in unilateral Below-Knee Amputation (BKA), therefore the aim of this study was to assess the interaction of knee, hip and L5-S1 joint contact forces between sound and prosthetic leg in patients with unilateral BKA during walking.

**Methods:** Five patients with BKA were recruited in this study. A Qualisys motion system captured with seven cameras and a Kistler force plate were used to record kinematics and kinetics variables of walking. The forces applied on knee, hip and L5-S1 joint contact forces (JCF) were calculated by using Open-SIM software. SPSS software was used to analyze data at an alpha set point of 0.05.

**Results:** The propulsive and second peaks of ground reaction forces applied on sound leg were significantly higher than on prosthetic leg ( $P < 0.05$ ). Although the forces applied on hip, knee and L5-S1 joint in the sound leg were higher compared to prosthetic leg, the interaction between side and joint factor was not significant ( $P > 0.05$ ).

**Conclusion:** The results of this study showed that the meaningful JCF applied on the sound leg was more than that of prosthetic leg. Insignificant increases in JCF on the sound leg during life can create cumulative forces on the knee and low-back and endanger these joints of the risk of knee OA and chronic low-back pain. Balanced forces applied on sound and prosthetic leg is important, if this is the case, so indicating using a proper application of socket pin and prosthetic feet may have beneficial impact on sound side loads.

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### Introduction

The prevalence of lower limb amputation is 2.8 to 43.9 individuals per 10,000 individuals in the United States [1]. The most common type of amputation is Below-Knee Amputation (BKA) [1].

In the BKA, the ankle and its muscular functions

are lost. The Solid Ankle-Cushion Heel (SACH) foot is used to bear the body weight during stance phase of gait. The disadvantage of this foot is its lack of flexibility. The mechanical dysfunctions of the SACH foot result in changes in kinetics and kinematics of walking in patients with BKA [2]. In order to perform daily tasks, patients need to be compatible with the prosthetic foot [3] so that the performance of their sound leg will increase so as to compensate dysfunctions of the prosthetic leg [4]. Otherwise, the incompatibility of the patient and prosthetic foot will lead to asymmetries of gait.

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There are many asymmetries during gait in patients with BKA including: The increment of stance phase in sound leg [3], the increment of prosthetic leg's step length [3, 4], the increment of Ground Reaction Force (GRF) applied on sound foot [5], abnormal muscle spasm in Medio-Lateral (M/L) and Antero-Posterior (A/P) planes [6] and lateral bending of trunk toward prosthetic leg [7]. These asymmetries of gait in patients with BKA change the pattern of loading applied on the back and lower limbs joints. Nolan *et al.* reported the prevalence of gait asymmetries to be under 10% in healthy individuals and over 23% in patients with amputation [8]. Other researchers have demonstrated that the symmetry index of gait has significant differences between healthy individuals and patients with BKA [9]. Gait asymmetry in patients with amputation can lead to joint pain and degeneration. Struyf *et al.* found that hip osteoarthritis in the sound limb of patients with amputation was 3 times greater than that of normal subjects [10]. Moreover, the increased mechanical work of extensor muscles in individuals with amputation increases the A/P force on hip joint in the sound leg [5].

Leg length discrepancy can cause low-back pain [11-13]. Walking with altered trunk kinematics in lower limb amputation patients create more compression on facets and tissues and can cause low-back pain [11, 14, 15]. Asymmetric gait results in the production of asymmetric forces on the joints leading to a faster degeneration of joints' cartilage [16].

Some investigators showed that peaks of vertical components of (GRF) [17] and loading rate [18] were higher in the intact limb of subjects with BKA compared to the contralateral side. In contrast, others demonstrated that the forces [19, 20], moment [20] and Joint Contact Forces (JCFs) [21] acting across the joints of the contralateral side were the same as those of normal matched subjects without disability. Previous studies indicated that intact sides of BKA have shown the incidence of knee and hip joint OA [10, 22-25]. It is therefore inferred that the increasing applied forces on intact limb may increase the incidence of OA, thus reduction of the forces applied on intact limb should be considered. If this is the case, it is important to investigate the magnitude of the forces applied on back and lower limb joint during walking in patients with BKA.

Previous research studies have examined the kinematics and kinetics of gait in BKA. To the best of our knowledge, there are few studies focusing on the forces applied on back and lower limb joint simultaneously [2, 5, 21, 26-28]. These studies assess the applied forces on back during sitting and standing up [2] or focusing on GRF and moments applied on low-back [27]. Although one study investigates the joint forces applied on low-back by calculating inverse dynamic [5], and one study uses the Open-SIM software to calculate JCFs in BKA for the first time, others assess the applied forces on low-back and lower limbs joints simultaneously [2, 5, 26-28].

In this study, Open-SIM software was used to calculate the JCF applied on knee, hip and L5-S1 joints. This study also demonstrates the pattern of forces applied on these

joints during walking. Understanding the patterns of applied forces on these joints via sound and prosthetic foot can be beneficial in clinical decision making. Therefore, the aim of this study was to assess interaction of knee, hip and L5-S1 JCFs in patients with unilateral BKA during walking. We hypothesized that the joint kinematics and the JCF of the sound side is different from the prosthetic side of subjects with BKA.

## Methods

Five men with BKA and a mean  $\pm$  D of age of  $48.75 \pm 3.77$  years, weight of  $732.6 \pm 67.62$  N and height of  $172.0 \pm 2.74$  cm were recruited in this study. The inclusion criteria were: 1) Not performing surgical procedures on legs affecting gait and independent walking and 2) using SACH foot with Modular prosthesis with polyfoam liner. Lack of stability during walking and reluctance to continue participation in the study were considered as exclusion criteria. An ethical approval was obtained from Ethical committee of Isfahan University of Medical Sciences. Moreover, each subject was asked to sign a consent form before data collection.

## Equipment

A Qualisys system with seven cameras was used to record kinematics variables of walking. A kistler force plate (500\*600 mm, 9260 AA model) (kistler company Switzerland) was used to record the GRFs during gait. Thirty six markers (14 mm diameter) were attached on the anterior superior iliac spine, posterior superior iliac spine, medial and lateral epicondyles, medial and lateral ankle malleolus, heel, 1<sup>st</sup> and 5<sup>th</sup> metatarsus, and left and right acromioclavicular joint. Four cluster markers were also attached on the left and right side of the leg and thigh. Figure 1 shows the marker position on the body based on the protocol approved by the Strasclde



Figure 1: A built model in Open-SIM software.

University [29]. The frequency of data collection was 120 Hz. Data were filtered with the frequency of 10 Hz [30]. Open-SIM software (Version 3.3, Stanford, USA) was used to model the patients' musculoskeletal system. The Open-SIM software is a simulator that can simulate JCF, muscle force and muscle length. It is also possible to distinguish between the pathologic and normal pattern with this software [31]. The data output of the Qualisys software were converted to be used in Open-SIM (Version 3.3) by Mokka software (powered by the open-source library Biomechanical ToolKit (BTK). Subjects' gait was modeled by using the lower extremities musculoskeletal model with a 23 degree of freedom and 92 muscles from Open-SIM (Version 3.3). All subject data were scaled using static trial data. Inverse kinematics, inverse dynamics tools and a residual reduction algorithm (RRA) tool were applied by using walking trials. The computer muscle control tool [32] was used to compute muscle excitations. Analyze tools were the last step to calculate JCF. JCFs are the sum of joint reaction forces and muscle tension applied on a joint (Eq. 1) (Figure 2)

$$F_{mus} + GRF + JCF_{dist} + JCF_{prox} = M_{seg} \cdot A_{seg}$$

Eq. 1

Where  $F_{mus}$  are the muscle forces (N) applied on the distal joint, GRF is the ground reaction force (N),  $JCF_{dist}$  is the compressive joint contact force (N) applied on the distal joint,  $JCF_{prox}$  is the joint force (N) applied on the proximal joint,  $M_{seg}$  is the mass of the segment, and

$A_{seg}$  is a 6 dimensional vector of rotational and translational acceleration of the segment [33, 34]

*Parameters*

After marker placement, each subject walked in the

walkway. Subjects walked for 5 times and the mean value of 5 trials were used for statistical analysis. There is evidence showing that a 5-trial set is an acceptable criteria for assessing the kinetics and kinematics variables [35, 36] of spatiotemporal parameter including: Cadence, walking speed, stride length, leg length, the peak GRFs in M/L, first and second peaks of GRFs and peak of braking and propulsive GRFs; JFCs of knee, hip and L5-S1 in three planes were evaluated in this study. Moreover, the range of motion of knee, hip, pelvis and back joints was reported in this study. Leg length was measured from the distance between greater trochanter marker to lateral malleoli when subjects stand on their foot in static position.

*Statistical Analysis*

The peaks of GRF and JCF components were normalized to body weight (BW). Normal distribution of data was confirmed by using the Shapiro-Wilk test. The independent t-test and the repeated measure test were used to analyze data. Repeated measure test was used to explore the interaction of side and joint factors. SPSS statistical software was used to analyze the data. The level of significance was set at 0.05.

**Results**

The mean values of anthropometric data of the patients with BKA are shown in Table 1.

The spatiotemporal parameters in subjects with BKA during walking

Table 2 shows the spatiotemporal variables of the BKA group during walking. There were no significant

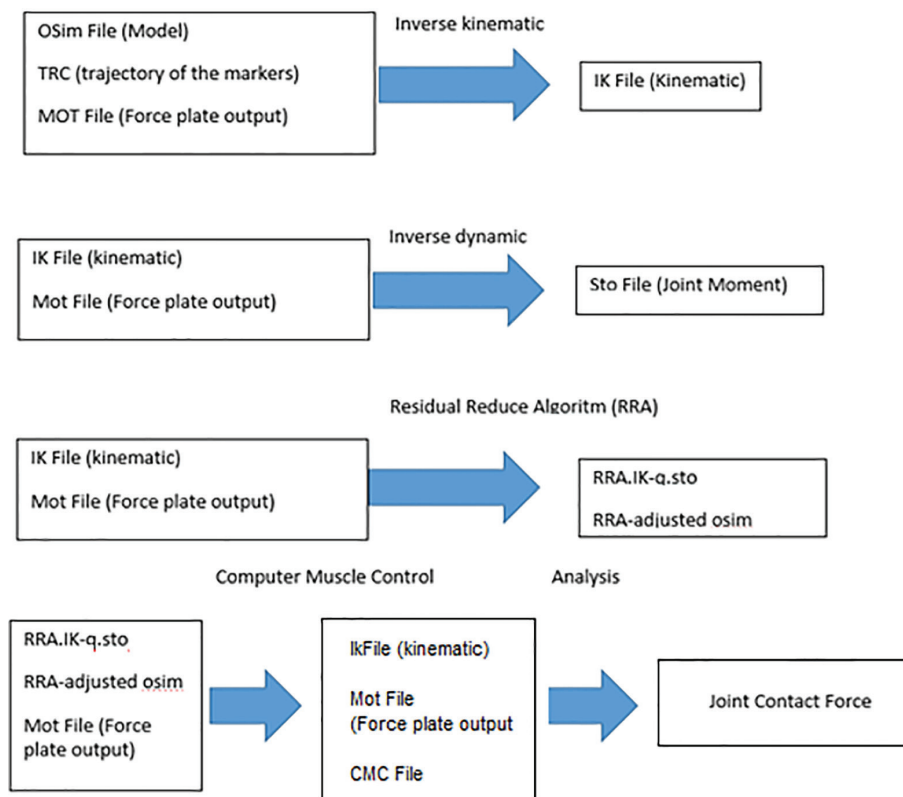


Figure 2: The procedure used in Open-SIM software to determine joint contact forces.

**Table 1:** The demographic information of subjects with BKA

Variables	Mean±SD
Age (year)	48.75±3.77
Weight (N)	732.60±67.62
Height (cm)	172.00±2.74
BMI (kg/m <sup>2</sup> )	24.71±2.49
Sound leg (cm)	88.80±0.85
Prosthetic leg (cm)	88.60±1.80
Pelvis length (cm)	36.80±2.50

**Table 2:** The mean and SD of the spatiotemporal parameters in subjects with BKA during walking

Variables	Sound leg Mean±SD	Prosthetic leg Mean±SD	t	P value
Stride time(s)	1.31±0.05	1.29±0.02	0.575	0.586
Stance (%)	63.83±1.97	69.58±9.39	-1.199	0.276
Swing (%)	36.17±1.96	31.42±10.09	0.925	0.391
Stride length(m)	1.28±0.07	1.31±0.07	-0.642	0.544
Speed (m/s)	0.98±0.08	1.01±0.07	-0.638	0.547
Cadence (step/min)	91.77±3.52	92.79±1.29	-0.549	0.603

differences in spatiotemporal variables between sound leg and prosthetic leg ( $P>0.05$ ). In prosthetic leg, the stride length ( $P=0.59$ ), speed ( $P=0.55$ ) and cadence ( $P=0.60$ ) were lower than that of sound leg by 0.2 (m), 0.03 (m/s) and 1(step/min), respectively. The percent of stance ( $P=0.27$ ) and swing ( $P=0.39$ ) phase of gait was the same between sound and prosthetic leg.

**Range of Motion of Knee, Hip, and Pelvis Joints During Walking**

Table 3 shows the range of motion of knee, hip, and pelvis joints during walking in sound and prosthetic legs. There were no significant differences between knee, hip, and pelvic range of motion ( $P>0.05$ ). Although the range of motion of hip (in all planes) and knee (in M/L and horizontal planes) joints was lower in prosthetic

leg compared to sound leg, these differences were not statistically significant. The highest differences between sound and prosthetic leg were seen in hip and knee joints range of motion by 5.4 and 17 degree, respectively.

**GRFs Applied on Sound and Prosthetic Leg During Walking**

The results of GRFs applied on the sound and prosthetic legs are shown in Table 4. The results indicated that the first peak ( $P=0.28$ ), braking ( $P=0.10$ ) and M/L ( $P=0.17$ ) of GRFs were not significant between sound and prosthetic legs. However, there were significant differences in the second peak ( $P<0.01$ ) and propulsive ( $P=0.05$ ) of GRFs applied on sound and prosthetic legs. The second peak propulsive component of GRFs applied on the sound leg were significantly higher than in prosthetic leg.

**Table 3:** The range of motion of knee, hip, and pelvis joints during walking (degree)

		Prosthetic leg Mean±SD	Sound leg Mean±SD	t	df	P value
Pelvis	X	5.54±0.26	5.59±1.84	-0.043	4	0.967
	Y	11.39±4.31	9.57±3.89	0.629	4	0.553
	Z	12.59±5.23	11.03±5.19	0.425	4	0.686
Hip	X	30.85±8.93	25.45±20.83	0.476	4	0.651
	Y	18.37±17.69	16.73±16.82	0.135	4	0.897
	Z	14.63±3.96	12.24±6.05	0.661	4	0.533
Knee	X	45.77±19.47	28.71±24.37	1.094	4	0.316
	Y	21.25±27.80	25.39±26.87	-0.214	4	0.837
	Z	15.37±4.02	14.26±9.49	0.215	4	0.837

X=Sagittal plane; Y=Frontal plane; Z=Horizontal plane

**Table 4:** The mean and SD of the ground reaction forces applied on sound and prosthetic leg (N/BW)

Variable	Prosthetic leg Mean±SD	Sound leg Mean±SD	t	P value
First peak (vertical)	1.05±0.10	1.12±0.10	-1.17	0.28
Second peak (vertical)	0.24±0.02	1.01±0.09	-18.58	0.00*
Anterio-posterior (braking) Force	0.10±0.06	0.15±0.03	-1.89	0.10
Anterio-posterior (Propulsive) Force	0.14±0.03	0.18±0.03	2.28	0.05*
Medio-Lateral Force	0.08±0.01	0.09±0.02	1.51	0.17

Show significant difference

*The JCF Production of Lower Limbs During a Gait Cycle*

Figure 3 shows no significant differences between the JCF applied on sound and prosthetic legs. Although there were no significant differences between the forces applied on sound and prosthetic leg ( $P=0.269$ ), the forces applied on sound leg were higher than that of prosthetic leg. Forces applied on sound leg were  $1.72\pm 0.18$  (N/BW) which was 0.31 (N/BW) higher than that of prosthetic leg.

As shown in Figure 4, there was no significant interaction between side and joint factors ( $P>0.05$ ). As can be seen in this graph, the forces applied on sound hip, knee and L5-S1 were higher than that of prosthetic leg but the differences were not significant.

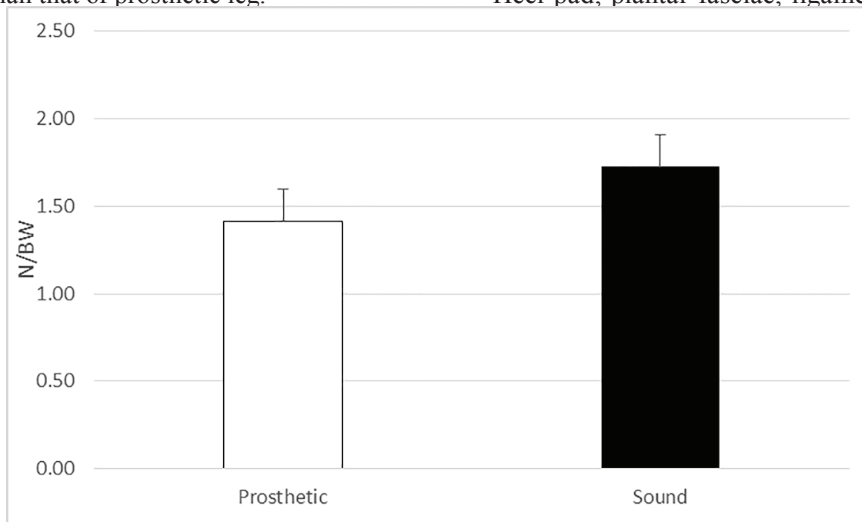
**Discussion**

The aim of this study was to assess the interaction of knee, hip and L5-S1 JCFs between the sound and prosthetic leg in subjects with unilateral BKA during walking. The result of the current study showed that there was no significant interaction between the JCFs pattern of hip, knee and L5-S1 joint in sound and prosthetic leg but the forces applied on knee and hip joints in the sound side were higher than that of prosthetic leg.

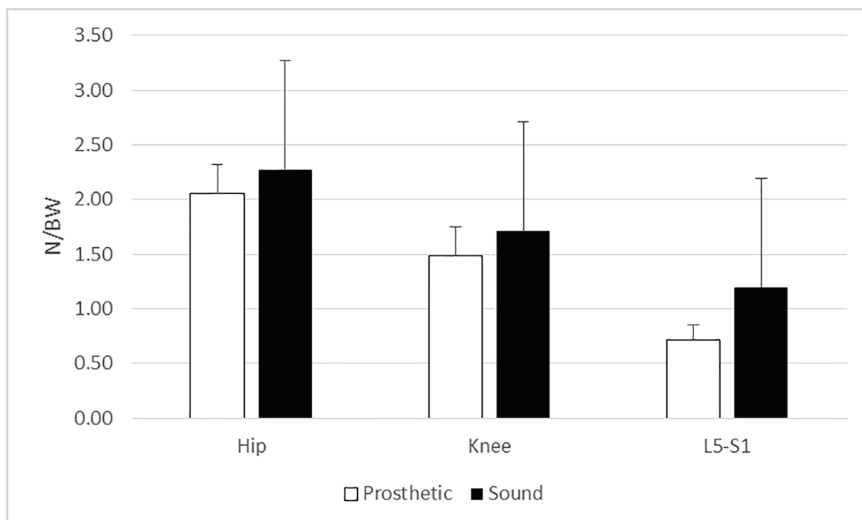
The results of GRFs also showed that greater forces are applied on the sound leg compared to prosthetic leg. These results are consistent with the results obtained by Sawaga *et al.* The results also showed no significant differences between walking speed, stride length and cadence in the sound leg compared to prosthetic leg. Sawaga *et al.* showed that subjects with BKA walked with less asymmetry compared with those with above knee amputation. Having intact knee is the main reason for less asymmetry during gait in BKA subjects [3]. Hendershot *et al.* reported that step length of the prosthetic leg was significantly greater than sound leg by 2 cm in patients with BKA, which is in contrast with the results of this study [27].

The results also revealed that the first peak of vertical, the braking and M/L GRF did not differ significantly between sound and prosthetic leg; however, there were significant differences in the second of vertical and propulsive GRFs applied on sound and prosthetic leg. The second peak of vertical and propulsive GRFs applied on sound side were significantly higher than in prosthetic leg. Beyaert *et al.* showed that GRFs applied on sound leg were significantly greater than prosthetic leg which is consistent with the results of this study [17].

Heel pad, plantar fasciae, ligaments and muscles in



**Figure 3:** Forces applied on sound and prosthetic legs



**Figure 4:** No interaction between side and joint factor

sound foot and ankle work normally during gait and act as a shock absorber during heel contact and act as a propeller during push off. These structures absorb the shock during heel contact and produce forces during push off in order to propel the body and help progression [37]. The highest pressure applied on foot during walking, is applied in heel contact phase and it can reach up to 130% of body weight [38]. Some of these forces are absorbed by heel pad [39], plantar fasciae and the concentric work of the soleus and eccentric work of tibialis anterior muscle [40] and the rest of the forces are translated to the knee joint

The SACH foot can only provide partial plantar flexion by some cushioning at the heel and would not be able to absorb the GRFs applied on this structure; however, it may be insufficient for a proper ROM, therefore, the SACH foot cannot act as an intact ankle in heel contact phase. Moreover, the prosthetic foot would not be able to perform plantar flexion and propel the body in push off phase. These functional differences between the SACH foot and a normal ankle can make the subjects who are using protective mechanism to decrease the forces applied on the prosthetic leg.

Although there was no significant interaction between the side and the joint factor (Figure 4), the forces applied on the knee, hip and L5-S1 joints are higher in sound leg compared to prosthetic leg. The possible reasons for these differences could be; (a) the propulsive and second peak of vertical GRFs applied on prosthetic leg were significantly lower than that of sound leg. (b) Another reason could be the poor function of the prosthetic leg which did not absorb the partial forces which are easily absorbed by the intact ankle, and the forces are directly translated to the knee. (c) Subjects with BKA walked with full or near full extension in stance phase of prosthetic leg. However, in normal walking, the knee joint flexes from a full extension in the initial heel contact phase, which is accompanied by the eccentric work of quadriceps muscles, so that the speed and flexion of the knee joint can decrease [40]. In this phase, the quadriceps muscles work eccentrically and absorb partial forces which are translated to the knee joint. Therefore, with the different vertical JCFs, the sound leg absorbs more forces in the ankle and knee joints whereas the prosthetic leg absorbs fewer forces. Consequently, the forces applied on knee and hip joint of the sound leg would be greater than that of prosthetic leg. Ch Yu et al. showed that the forces applied on the joints of sound leg are greater compared to prosthetic leg which is consistent with the result of this study [5]. Leg length discrepancy may be another reason for different JCFs between the sound side and prosthetic side. During walking, there is an increase in the rate and magnitude of impact loading on the intact limb, whereas loading of the prosthetic limb is actually less than normal [22-24]. Prosthetic foot design can influence the abnormal loading characteristics of the intact limb. Asymmetric gait due to leg length discrepancy can apply asymmetric forces on the joints leading to faster degeneration of the joints cartilage [16]. Walking with leg length discrepancy and JCFs between sound and prosthetic leg are related to the kinematics of pelvis and thorax motion. The thorax and

pelvis motion are in-phase during gait in patients with BKA [41]. That is, during the stance phase of sound leg, the contralateral pelvis goes higher than the ipsilateral pelvis [42] and the thorax deviates to prosthetic leg in A/P plane [43]. This altered kinematics of pelvis and thorax lead to asymmetrical loading on L5-S1 joint and can lead to muscle spasm and chronic low-back pain altered trunk kinematics in lower limb amputation patients, thus creating more compression on facets and tissues and can cause low-back pain [11].

The JCFs applied on L5-S1 joint were greater in stance phase of sound leg compared to prosthetic leg. Ch Yu et al. showed that the forces applied on the back and on the sound leg were significantly greater than that of prosthetic leg which is consistent with the result of this study [5] and is the possible reason for the differences in L5-S1 in subjects with BKA [44].

#### Limitation

Small sample size of the subjects is one of the limitations of this study; therefore, future studies should be done with a large sample size. It is proposed that future studies assess the effect of SACH foot on the back and trunk muscle activation and forces during walking. Another limitation was lack of adequate control over the participants. That is, the inclusion criteria were not enough because any existence of pressure sores, pain, joint contractures, and spasm needed to be checked before participation. Moreover, experience of prosthesis use needed to be controlled.

#### Conclusion

The results of this study showed that the meaningful JCF applied on the sound leg was more than that of prosthetic leg. Insignificant increases in JCF on the sound leg during life can create cumulative forces on the knee and low-back and endanger these joints of the risk of knee OA and chronic low-back pain. Balanced forces applied on sound and prosthetic leg is important, if this is the case, so indicating using a proper application of socket pin and prosthetic feet may have beneficial impact on sound side loads.

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**Conflict of interest:** None declared.

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