The Journal of Foot & Ankle Surgery ■■ (2017) ■■–■■

Contents lists available at ScienceDirect

The Journal of Foot & Ankle Surgery

journal homepage: www.jfas.org

Original Research

Dynamic Functional Stiffness Index of the Ankle Joint During Daily Living

Hande Argunsah Bayram, PhD [1,2], Mehmed B. Bayram, PhD [1,2]

[1] *Assistant Professor, Department of Medical Engineering, Acibadem Mehmet Ali Aydinlar University, Istanbul, Turkey*
[2] *Doctoral Student, Department of Biomedical Engineering, Lerner Research Institute, Cleveland Clinic, Cleveland, OH*

ARTICLE INFO

Level of Clinical Evidence: 1

Keywords:
amputation
ankle
prosthetics
stiffness
transtibial

ABSTRACT

Exploring ankle joint physiologic functional stiffness is crucial for improving the design of prosthetic feet that aim to mimic normal gait. We hypothesized that ankle joint stiffness would vary among the different activities of daily living and that the magnitude of the stiffness would indicate the degree of energy storage element sufficiency in terms of harvesting and returning energy. We examined sagittal plane ankle moment versus flexion angle curves from 12 healthy subjects during the daily activities. The slopes of these curves were assessed to find the calculated stiffness during the peak energy return and harvest phases. For the energy return and harvest phases, stiffness varied from 0.016 to 0.283 Nm/kg° and 0.025 and 0.858 Nm/kg°, respectively. The optimum stiffness during the energy return phase was 0.111 ± 0.117 Nm/kg° and during the energy harvest phase was 0.234 ± 0.327 Nm/kg°. Ankle joint stiffness varied significantly during the activities of daily living, indicating that an energy storage unit with a constant stiffness would not be sufficient in providing energy regenerative gait during all activities. The present study was directed toward the development of a complete data set to determine the torque-angle properties of the ankle joint to facilitate a better design process.

The number of people living with a limb amputation is approximately 2 million (1), with nearly 185,000 amputations occurring in the United States every year (2). Transtibial amputation constitutes 53% of the total incidents. Diabetes and vascular diseases are the foremost causes of amputations, accounting for the 33% of the overall cases. The remaining cases result from tumor, trauma, and infection (3). Without effective interventions to prevent diabetes and vascular diseases and their complications, the potential is high for the number of amputations to increase. This enhances the importance of developing adaptive prosthetics capable of mimicking healthy ambulation (4), because inadequate prosthesis use, not only reduces the quality of life, but also accelerates other metabolic and cardiovascular problems.

Transtibial amputation is one of the most traumatic treatments a patient can receive. Amputees not only lose physical function and neural feedback after amputation but also have altered lower limb kinematics, which causes abnormal pelvic movements in the frontal plane (5). Amputees also have reduced neuromuscular feedback and control that diminishes their sense of balance and increases the probability of falls (6). Improper pelvic movement is another consequence of amputation and results in some pathologic problems such as osteoarthritis and lower back pain (7).

When a lower extremity joint harvests energy, it absorbs energy from the lower extremity. When this absorbed energy is needed to maintain the balance and safe mobilization, the joint generates the right amount of it to the tissues. The ankle acts as an energy generator during the preswing phase and stores 80% of the needed energy during walking at the self-selected walking speed (8). Therefore, the energy flow at the ankle is highly significant. During the stance phase, the knee harvests energy during flexion and returns as much as the same amount for the rest of the stride. In contrast, the ankle harvests energy as a result of weightbearing during this period. During the preswing phase, the knee continues harvesting energy, and the ankle returns a significant amount of it for the push-off. This energy flow occurs through the gastrocnemius and plantaris muscles. Therefore, a sufficient amount of energy for the knee is obtained and, thus, an assisted stride is achieved. As such, if a transtibial prosthesis provides correct positioning of the foot throughout the stride, it can help restore a physiologic gait pattern. Additionally, if it is supplemented with an appropriately designed spring, it can provide the required dorsiflexion, with hard stops at the limits of the range of motion.

Financial Disclosure: This study was funded by the Ohio Department of Development, State of Ohio (TECH 09-001), which provided funding in support of the Rapid Rehabilitation and Return to Function for Amputee Soldiers project.

Conflict of Interest: None reported.

Address correspondence to: Hande Argunsah Bayram, PhD, Department of Medical Engineering, Acibadem Mehmet Ali Aydinlar University, Kerem Aydinlar Campus, Icerenkoy Mah. Kayisdagi Cad. No:32 Atasehir, Istanbul, Turkey.

E-mail address: hande.argunsah@gmail.com (H. Argunsah Bayram).

https://doi.org/10.1053/j.jfas.2017.11.034

Table 1
Participant physical and demographic information

Subject No.	Sex	Age (y)	Height (cm)	Weight (kg)	BMI (kg/m^2)	Neuropathy	Race
1	Male	25	184.1	91.4	27.0	None	White
2	Male	28	171.4	71.2	24.2	None	White
3	Female	23	161.3	58.0	22.3	None	White
4	Male	29	176.5	92.8	29.8	None	White
5	Female	37	171.4	79.5	27.1	None	White
6	Female	32	163.8	58.4	21.8	None	White
7	Female	25	156.8	45.4	18.5	None	Asian
8	Male	36	188.0	80.4	22.7	None	White
9	Female	38	150.5	49.8	22.0	None	White
10	Female	29	163.8	62.0	23.1	None	Black
11	Male	32	185.4	85.6	24.9	None	Asian
12	Male	31	180.3	74.7	23.0	None	Black

Abbreviations: BMI, body mass index; No., number.

Because gait is a cyclical pattern that has positive and negative work phases, using a spring can reduce the power demand significantly by imitating the musculotendinous structures and thus achieve a high efficiency and power/weight ratio (9). Theoretically, energy is harvested from the weight of the body during the initial impact and then released at push-off, which makes the spring the main source of energy regeneration. The stiffness of the ankle joint during walking at variable cadences has been investigated by Hansen et al (10). However, other activities that demand more positive energy such as stair walking and running were not analyzed. In another recent study, ankle stiffness was investigated during quiet standing. Postural sway was analyzed by calculating the sway angles and sway moments of force related to both mediolateral and anteroposterior sway directions (11,12).

Measuring ankle stiffness is crucial for improving the design of transtibial prostheses that aim to imitate the normal gait. We hypothesized that the ankle joint stiffness would vary among the different activities of daily living. Therefore, a transtibial prosthesis should be adaptive to mimic ankle movement during different activities. Moreover, although the ankle performs natural movements, the magnitude of the nTC (mean averaged joint torque capacity per joint rotation degrees) indicates the degree of energy harvest element sufficiency of harvesting and returning energy when needed. The present investigation aimed to understand the energy flow at the ankle joint to guide the prosthetic device manufacturers theoretically. We randomly selected 12 healthy adults who met the study inclusion criteria and collected ankle joint biomechanical data during the dynamic conditions of daily living.

Patients and Methods

Participants

The aim of the present study was to collect the biomechanical data from randomly selected healthy adults to determine the ankle joint energy flow during activities of daily living. Using local advertisements, we recruited 12 healthy volunteer adult participants (6 males) aged 23 to 38 years (mean 30 ± 4 years; mean height 1.71 ± 0.05 m; mean mass 70.8 ± 15.9 kg; mean body mass index [BMI] 23.9 ± 2.9 kg/m^2) who did not have neuropathy or any obvious gait abnormality (Table 1). Only normal weight people, whose BMI was within 18.5 to 30 kg/m^2 (available at: www.heartfoundation.org), were recruited for the present study. The ratios of the ethnic categories were calculated according to the U.S. demographic 2011 database. The data collection of the participants started in August 2012 and ended in November 2012. Before testing, each participant read and signed an informed consent form that had been approved by the institutional review board.

The inclusion criteria were age 18 to 45 years, no gait abnormalities, and a normal BMI. The exclusion criteria were pregnancy, previous lower extremity surgery, and the presence of neuropathy.

The investigated parameters had a standard deviation of 10% to 15% of the mean. With a sample size of 12, we had 95% confidence that the population mean would be within approximately ±5% of the sample mean.

Methods

The motion analysis system consisted of an 8-camera motion capture system (Motion Analysis Corp., Santa Rosa, CA) and an AMTI force plate (AMTI Force and Motion, Watertown, MA). Each participant was set up with 34 retroreflective markers (Motion Analysis Corp.) using the Cleveland Clinic marker set. The Cleveland Clinic marker set attached the markers to represent joints and specific body landmarks. All participants walked wearing soft-soled gym shoes. Before data collection, 2 standing collections were taken to confirm the visibility of all markers within the system. Three acceptable walking collections were taken before data collection to obtain a consistent locomotion rhythm during walking at variable cadence and slow running activities.

A standard testing protocol was developed, and the same routine was followed for each participant. The study paradigm included a total of 6 daily activities: slow walk, normal walk, fast walk, slow run, step up 18-cm-high and 31-cm-deep steps, and step down the same steps (Table 2). The velocity was calculated for the walking and running activities. The pathway was 10 m long, and the elapsed time was determined using a stopwatch. Five acceptable trials of the foot squarely striking the force plate were collected and averaged to represent each collected parameter. A trial was considered acceptable as long as the participant had appropriately stepped on the force plate with the intended foot while maintaining the desired ambulation rhythm for the walking and running trials. The normal walking speed was considered the self-selected speed. Before data collection, the time required was given to participants to allow them to practice and become used to maintaining the ambulation rhythm during the walking and running activities. The slow and fast walking trials were defined as 25% slower and 25% faster than the normal walking speed, respectively. After achieving a consistent slow/fast walking rhythm for each participant, data collection was initiated. During slow running activity, each participant was asked to run at their self-selected pace.

For the stair ascent/descent trials, full contact with the force plate with the intended foot was defined as an acceptable trial. A 3-step staircase was placed on the force plate during these activities. The participants were asked to climb up and down to the steps, which had been placed over the force plate. The second step up/down was used in the calculations, because it was considered the perfectly balanced step.

Table 2
Spatiotemporal measurements across activities of daily living (N = 12)

Variable	Stair Ascent	Stair Descent	Slow Walk	Normal Walk	Fast Walk	Slow Run
Cadence (steps/min)	75.2 ± 4.5	86.3 ± 7.5	101.9 ± 7.3	113.9 ± 7.6	125.3 ± 11.8	157.9 ± 10.1
Step length (cm)	26.8 ± 7.3	32.2 ± 7.8	63.4 ± 8.5	72.1 ± 8.4	78.2 ± 9.8	91.8 ± 12.3
Step width (cm)	10.6 ± 4.5	12.1 ± 4.7	12.0 ± 2.6	12.5 ± 2.5	11.9 ± 2.5	11.6 ± 2.9
Forward velocity (cm/s)	40.4 ± 4.1	44.1 ± 8.3	108.4 ± 16.3	138.4 ± 18.4	166.6 ± 24.7	242.7 ± 33.2

Data presented as mean ± standard deviation.