Clinical Biomechanics 29 (2014) 1193–1199

Contents lists available at ScienceDirect

Clinical Biomechanics

journal homepage: www.elsevier.com/locate/clinbiomech

Lecture

Active functional stiffness of the knee joint during activities of daily living: A parameter for improved design of prosthetic limbs

Hande Argunsah Bayram [a,b,c,d,*], Chih-Hao Chien [a,b], Brian L. Davis [e]

[a] Department of Chemical and Biomedical Engineering, Cleveland State University, Cleveland, OH, USA
[b] Department of Biomedical Engineering, Lerner Research Institute, Cleveland Clinic, Cleveland, OH, USA
[c] Human Performance and Engineering Laboratory, Kessler Foundation, West Orange, NJ, USA
[d] Department of Research, Children's Specialized Hospital, New Brunswick, NJ, USA
[e] Department of Biomedical Engineering, University of Akron, Akron, OH, USA

ARTICLE INFO

Article history:
Received 22 April 2013
Accepted 7 September 2014

Keywords:
Prosthesis design
Healthy human knee
Physiological functional stiffness

ABSTRACT

Background: Exploring knee joint physiological functional stiffness is crucial for improving the design of prosthetic legs that aim to mimic normal gait. This study hypothesizes that knee joint stiffness varies among different activities of daily living, additionally while the knee performs natural movements; the magnitude of the stiffness indicates the degree of energy storage element sufficiency in terms of harvesting/returning energy.
Methods: This study examined sagittal plane knee moment vs. knee flexion angle curves from 12 able-bodied subjects during activities of daily living. Slopes of these curves were assessed to find the calculated stiffness during the peak energy return and harvest phases so that the activities, which can be performed when the prosthetic knee is supplemented by a spring, were identified.
Findings: For the energy return and harvest phases, the stiffness varied between 0.006 and 0.046 Nm/kg deg. and 0 and 0.052 Nm/kg deg. respectively. The optimum energy return phase stiffness was 0.024 (SD 0.013) Nm/kg deg. and energy harvest phase stiffness was 0.011 (SD 0.018) Nm/kg deg.
Interpretation: Knee joint stiffness varied significantly during activities of daily living, which indicated that a storage unit with a constant stiffness would not be sufficient in providing energy regenerative gait during all activities. However, by controlling the amount and timing of spring compression and release, an energy-regenerative prosthetic knee device could be developed during most of the activities. This study was directed to the development of a complete data set, which determined the torque-angle properties of the healthy knee joint.

1. Introduction

The prosthetic knee market provides the amputee with many choices including purely passive, hydraulic, pneumatic, friction-based and powered models. Current generation prostheses, except the purely passive models, are supplemented by microprocessors and a variety of sensor techniques such as electromyography, knee angle sensors, heel strike/toe off indication switches and foot load cells. Nevertheless, the gait quality of transfemoral amputees has not improved significantly in the last 50 years due to the inefficient knee joint energy flow distribution of the commercial prosthetic knees along with the absence of the proper knee flexion during the swing phase, causing the patient to hike his/her pelvis, which leads to back pain in the long run. In this context, state-of-art prosthetics technology is trending toward creating energy regenerative devices, which are able to harvest and return metabolic energy during ambulation. Power Knee is an energy regenerative prosthesis however it is rarely used since it has not been given a Medicare insurance reimbursement code due to its high price and reports on its performance are scarce in the literature. Currently only two Iraq war veterans are using the Power Knee (Dunne, 2012). Besides its almost triple total weight when it is compared with the commercial passive and microprocessored designs, it can only be used by the unilateral transfemoral amputees as in order to mimic the movement of the able-bodied human ambulation by producing the required positive energy, the sensor on the amputated limb needs the real time data collected by the sensor on the intact limb. Among the people who live with limb loss, the fraction of bilateral transfemoral amputees is approximately 18% (Stewart and Jain, 1993). In addition to the significantly high market price drawback of Power Knee, when the dramatically increasing obesity and diabetes rates are considered, it is anticipated that the number of bilateral transfemoral amputation will increase, that will reduce the number of potential Power Knee users.

There are several commercial systems that can provide positive energy to the amputees through motors. Although these systems have

* Corresponding author at: Human Performance and Engineering Laboratory, Kessler Foundation, 1199 Pleasant Valley Way, West Orange, NJ 07052, USA.
E-mail address: hande.argunsah@gmail.com (H. Argunsah Bayram).

http://dx.doi.org/10.1016/j.clinbiomech.2014.09.004

many drawbacks such as significantly heavy components that reduces the power-to-weight ratio, limited battery life and extremely hot hardware after ambulation, they are energy regenerative and allow amputees to ambulate more naturally compared to semi-active prosthetic systems during more energy demanding activities of daily living (ADL) such as stair climbing. This study identifies the activities, which do/do not require positive energy and aims to utilize energy flow and/or control stiffness parameters to make the motored systems more efficient.

Energy flow deficiencies due to a lack of positive work during the stance phase of gait for patients using controlled damping prostheses, led developers to consider emerging designs with controlled energy harvest/return capabilities (Martinez-Villalpando and Herr, 2009; Sup et al., 2008; van den Bogert et al., 2012). The use of a spring in a prosthetic knee device theoretically reduces power demands and also provides high power-to-weight ratio (Argunsah and Davis, 2012). Therefore, instead of using heavy motors, gearboxes and bulky batteries, a spring can help the peak power demand of the prosthesis during the stance phase of gait with less weight. Moreover, adverse consequences such as circumduction and a disturbed gait pattern that take place due to the absence of positive energy delivery during the swing phase could potentially be alleviated by an energy-harvesting and timed-release system. Current generation powered designs deliver the needed energy with heavy battery demanding motors, which increase the mass of the device significantly compared to the other commercial prosthetic knees (Power Knee weighs 4700 g, whereas the average weight of the commercial microprocessored knees is 1250 g), however, patients want lightweight prosthetics which are capable of delivering natural gait by providing normal knee and ankle flexion angles during the ADL (Aeyels et al., 1992). In essence, amputee patients prefer the prosthetic limb to be as light as possible — certainly lighter than their natural limb. This may be due to the fact that they have significantly weakened musculature so the prosthetic knee manufacturers aim to design prosthetic knees which are energy regenerative and at the same time as light as possible.

Constant spring stiffness is suboptimal to varying gait requirements for different types of motion, due to the variability of the impedance (Pfeifer et al., 2012), physiological functional stiffness (PFS) and the power requirement of the knee. Passive characteristics, viscous and elastic attributes and the activation dependent properties of the muscles and their surroundings in the joint (Johns and Wright, 1962; Winters et al., 1988) make energy regenerative prosthetic knee device with simple mechanisms challenging. These properties are altered instantaneously as the muscles are activated depending on the biomechanical demands of the performed activity and are closely linked to the instant stiffness of the joint. This adaptation is achieved through the active elasticity adjustment of the muscles in the joint. Thus, attaining this adaptation with controlled damper mechanisms, which are not capable of mimicking the activation dependent properties of the muscles and their surroundings in the joint, is hardly possible.

Passive stiffness, which is the measure of the flexion torque and the angle of the knee joint, during controlled perturbations without activating the muscles, has been investigated before (Mansour and Audu, 1986; Silder et al., 2007; Whittington et al., 2008; Yoon and Mansou, 1982; Zhang et al., 1998). Pfeifer et al. (2012) developed an active knee-stiffness analysis model; however, only level ground walking has been investigated in this study. Hansen et al. (2004) have examined the ankle joint active PFS during walking at variable cadence for guiding the prosthetic device designers in developing transtibial prostheses, which can optimize the energy flow at the ankle joint, yet, the authors are not aware of any prior work on the quantitative active knee joint PFS analysis across a wide range of daily activities.

Amputees not only lose physical function and neural feedback after amputation, but also have altered knee kinematics during the swing phase of the prosthetic limb, which causes abnormal pelvic movements in the frontal plane even with the most sophisticated microprocessored prosthetic knee available on the market today (Gailey et al., 2008; Kulkarni et al., 2005; Modan et al., 1998). Due to this fact, rather than modeling a prosthetic knee like an able-bodied knee, these factors and their impact on the amputee gait should be considered.

2. Methods

2.1. Subjects

Twelve healthy, relatively young female and male volunteer subjects (mean age: 30.42, SD: 4.87 yr.; mean height: 1.71, SD: 0.049 m; mean mass: 70.8, SD: 15.9 kg; mean BMI: 23.9, SD: 2.99 kg/m^2) who do not have neuropathy and did not show any obvious gait abnormalities, participated in this study. Only normal weight subjects participated as, in our opinion, those amputees who are most likely to benefit from these kinds of prosthetic systems are K3 and K4 patients, and these are typically not overweight individuals. Prior to testing, each subject read and signed an informed consent that was approved by the Cleveland Clinic Foundation Institutional Review Board. Spatio-temporal measures across the ADL were calculated (Table 1) for each ADL.

The peak energy-harvest/return phases were identified according to the magnitude of the total area under the power vs. percentage of stride curves (Fig. 1). The maximum positive work peak was identified as the R_{max} and the maximum negative work peak as the H_{max} for each activity (Table 2). The areas under the R_{max} and H_{max} curves were calculated by the following equations:

$$R_{max} = \max W_i^{positive} \quad \text{where} \quad W_i^{positive} = \int_{t_i}^{t_f} P_{d_t} \tag{1}$$

$$H_{max} = \max W_i^{negative} \quad \text{where} \quad W_i^{negative} = \int_{t_i}^{t_f} P_{d_t}. \tag{2}$$

2.2. Experimental protocol

The motion analysis system used in the Cleveland Clinic's “Doris Alburn Musculoskeletal Performance Laboratory” consisted of an eight-camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA) and an AMTI force plate (AMTI Force and Motion, Watertown, MA). Each subject was setup with 34 Motion Analysis Corporation retroreflective markers using the Cleveland Clinic marker set which attaches the markers to represent joints and specific bony landmarks (Fig. 1). Twelve of the markers were composed of triangle marker sets (19 mm) and placed on the right and left thighs and shanks of the subject. The remaining marker placements include the right and left shoulders, right and left elbows, right and left wrists, lateral and medial knees

Table 1
Means, SD of spatio-temporal measures across ADL (n = 12).

	Stair ascent	Stair descent	Sit/stand/sit sequence	Slow walking	Normal walking	Fast walking	Slow running
Cadence (steps/min)	75.2 (SD 4.5)	86.3 (SD 7.5)	35.5 (SD 4.2)	101.9 (SD 7.3)	113.9 (SD 7.6)	125.3 (SD 11.8)	157.9 (SD 10.1)
Step length (cm)	26.8 (SD 7.3)	32.2 (SD 7.8)	0.012 (SD 1.37)	63.4 (SD 8.5)	72.1 (SD 8.4)	78.2 (SD 9.8)	91.8 (SD 12.3)
Step width (cm)	10.6 (SD 4.5)	12.09 (SD 4.7)	28.1 (SD 7.3)	12.0 (SD 2.6)	12.5 (SD 2.5)	11.9 (SD 2.5)	11.6 (SD 2.9)
Forward velocity (cm/s)	40.4 (SD 4.1)	44.1 (SD 8.3)	0.65 (SD 0.4)	108.4 (SD 16.3)	138.4 (SD 18.4)	166.6 (SD 24.7)	242.7 (SD 33.2)