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Design and Validation of a Platform Robot for Determination of Ankle Impedance During Ambulation

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***Abstract*—In order to provide natural, biomimetic control to recently developed powered ankle prostheses, we must characterize the impedance of the ankle during ambulation tasks. To this end, a platform robot was developed that can apply an angular perturbation to the ankle during ambulation and simultaneously acquire ground reaction force data. In this study, we detail the design of the platform robot and characterize the impedance of the ankle during quiet standing. Subjects were perturbed by a 3° dorsiflexive ramp perturbation with a length of 150 ms. The impedance was defined parametrically, using a second order model to map joint angle to the torque response. The torque was determined using the inverted pendulum assumption, and impedance was identified by the least squares best estimate, yielding an average damping coefficient of 0.03±0.01 Nms/° and an average stiffness coefficient of 3.1±1.2 Nm/°. The estimates obtained by the proposed platform robot compare favorably to those published in the literature. Future work will investigate the impedance of the ankle during ambulation for powered prosthesis controller development.**

I. Introduction

THE ability to accommodate external disturbances is an essential part of posture, balance and movement. This is reflected by the dynamic relationship between the angular position of a joint and the corresponding torque generated, a concept known as joint impedance. There have been significant investigations into joint impedances in both the upper and lower-limbs, including the wrist [1], elbow [2], [3], knee [4] and ankle [5]-[9]. Specifically, measurements of ankle impedance have been determined for a broad array of static tasks, including sitting [10], laying supine [11], and standing [12], [13]. These interesting studies have shown that joint impedance varies with mean joint position [8], [9], neural activation [14], [15], perturbation amplitude [6], and applied torque [7], and have elucidated the level of neural control necessary for stable quiet standing [12], [13]. These properties are extremely relevant for postural control and static tasks, but provide little information about joint impedance when neural activation and/or joint position are changing, as is the case with many everyday activities.

With the advent of clinically viable powered ankle prostheses [16], [17], there is presently a need to develop natural, biomimetic control systems to optimally utilize this technology. Currently, powered ankle prostheses use impedance control, where the output torque is a function of angular displacement. The impedance control systems are tuned to match the "quasi-stiffness" of the ankle; a concept that was originally published by Hansen et al. for passive prosthesis design [18]. However, similarly to the human ankle, these prostheses can modulate their impedance in addition to emulating the quasi-stiffness. Hence, current powered prosthesis control results in an inaccurate representation of the natural impedance of the able-bodied ankle and the appropriate ankle impedance parameters are unknown.

The purpose of the current study is to detail the design of a platform robot, termed the Perturberator, which will eventually be used to determine the able-bodied ankle impedance during ambulation. Before the Perturberator can be used for this task, it must be validated to demonstrate its accuracy.

The Perturberator can perturb the ankle about its center of rotation during the stance phase of walking and record the ground reaction force data. To demonstrate the validity of the Perturberator, the impedance of the ankle during standing was determined using parametric least squares estimation techniques and subsequently compared to values published in the literature.

A. Previous Ankle Perturbation Devices

There have been many devices developed to perturb the ankle, both in a fixed posture and during walking [12], [13], [19]-[23]. The devices developed to perturb the ankle during walking typically span the ankle joint (i.e. ankle-foot orthosis type designs). These devices are usually designed to study reflex response to perturbations or aid rehabilitative training [19]-[22]. However, when attempting to identify the impedance of the ankle during walking with such a device, the measurement can be confounded by the device's own mechanical impedance. This creates a feedback loop that prevents accurate impedance identification. Additionally, the perturbation power requirements make it difficult to design a

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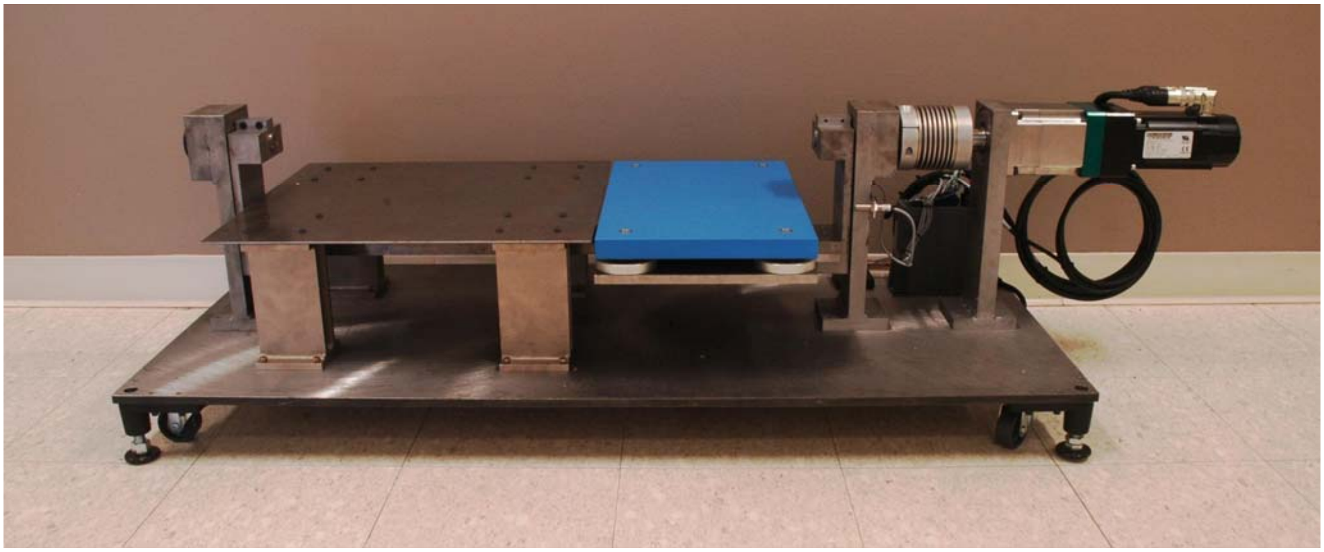

Fig. 1: The swing of the Perturberator spans the two hinge joints on either side of the device. The force platform is shown mounted to the swing; and the gearmotor and coupler can be seen on the right hand side. The Perturberator is built to span a walkway (not shown) such that subjects can be perturbed during ambulation.

device that is not so heavy it alters normal gait patterns. Other perturbation devices have been developed that do not span the ankle joint; however, their designs prevent them from being used to perturb the ankle during ambulation tasks [12], [13].

There have also been attempts to determine the ankle "stiffness" during dynamic activities without the use of a perturbation device. Such work focuses on stepping down and "bouncing gaits" including hopping and running [23]-[26]. Since the neuromuscular system is capable of introducing work into the system (unlike passive springs), these analyses cannot be used to estimate the actual "stiffness" or impedance. Instead, these results determine the torque-angle relationship necessary for the tasks. A similar analysis was used to determine the aforementioned quasi-stiffness of the ankle.

II. Description of the Ankle Perturbation Device

The Perturberator is a novel biomedical joint impedance identification device that allows for the perturbation of the ankle during the stance phase of walking and the acquisition of ground reaction force data. The device has a single degree of freedom (DOF), which is actively controlled. A motor drives the angle of a hinge-like swing that includes a mounted portable force platform. The center of rotation of the swing can be adjusted vertically by adding spacers (Fig. 1). To increase the overall stiffness of the device, all components were machined out of AISI 1020 steel. The platform is designed to span an adjustable aluminum walkway on either side (not shown), such that the platform is flush with the walkway.

The Perturberator uses a Kollmorgen (Radford, VA) AKM-42H brushless dc motor with a peak torque of 9.3 Nm and a rated power of 1.25 kW. The motor is powered by 120 V ac supply and fused at 15 A. The motor output is augmented by a gear reduction of 70:1, using a Kollmorgen ValueTRUE gearhead, resulting in an overall peak torque of 653 Nm. The gearhead output shaft is coupled to the swing via a KM-400 flexible bellows coupler (GAM, Mount Prospect, IL), with a rated torque of 400 Nm. The series torsional stiffness of the coupler and gearhead is 12 Nm/arc minute. The Perturberator has a maximum dorsiflexive and plantarflexive range of motion of 20°. The motor is controlled using a Kollmorgen AKD servo drive and a PIC32 microcontroller (Microchip, Chandler, AZ). The microcontroller outputs a "step and direction" paradigm that is mapped to angular velocity by varying the frequency of the pulse width modulation output. The servomotor is specified to have 20,000 steps per revolution and the microcontroller loop updates at approximately 5 kHz. Given the desired position input from the microcontroller, the servo drive closes the position, velocity and current loops. These

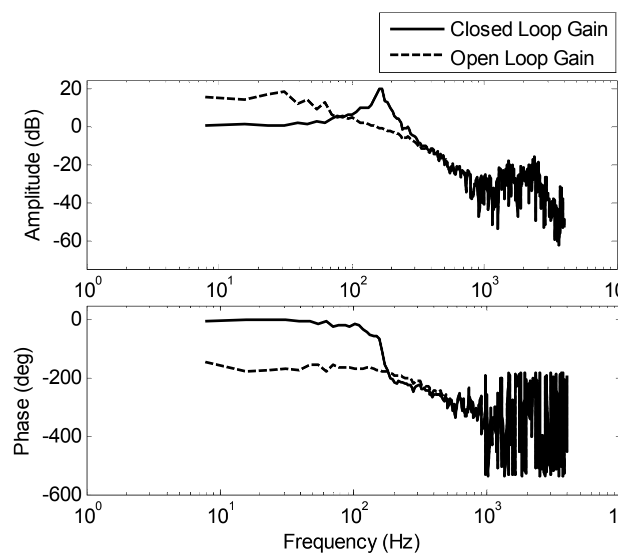

Fig. 2: Controller closed loop and open loop bode plots. Note the second order response of the closed loop transfer function with a natural frequency of approximately 200 Hz.

loops update at frequencies of 8 kHz, 16 kHz and 1.5 MHz, respectively. The closed loop frequency response has a natural frequency of approximately 200 Hz (Fig. 2), with a gain margin of 8 dB and phase margin of 45°. To ensure system safety, the Perturberator has two position-adjustable inductive limit switches, as well as an emergency stop switch and hard max-position stops.

The ground reaction force is measured using a Kistler 9260AA3 portable force platform (Winterthur, Switzerland) with a z-axis natural frequency of approximately 300 Hz and an x- and y-axis natural frequency of approximately 500 Hz. The drift associated with this force platform is less than 10 mN/s. The signals are acquired using a PC with Matlab (The Mathworks, Natick, MA) and a 16-bit National Instruments (Austin, TX) USB-6218 data acquisition card.

III. Methods

A. Experimental Protocol

Four healthy subjects, three male and one female, took part in this preliminary study. The subject's ages ranged from 24 to 30. Subjects gave written informed consent and the study was approved by the Northwestern University Institutional Review Board.

The center of rotation of the Perturberator swing was adjusted to the approximate height of the subject's ankle center of rotation. Subsequently, subjects stood barefoot with both feet on the force platform and the centers of rotation of their ankles were visually aligned with the center of rotation of the Perturberator swing [27]. Subjects stood quietly with their feet shoulder width apart.

The Perturberator applied a 3° dorsiflexive ramp perturbation that lasted 150ms. The magnitude of the perturbation was chosen to be similar to the disturbances encountered during walking tasks. Subjects were perturbed in two trials of 11 perturbations each. Between each perturbation, there was a waiting period that randomly varied between 10 seconds and 1 minute.

Data acquired included force platform information, motor angle, and ankle angle relative to shank, all sampled at 1 kHz. The motor angle was output from the servo drive with an angular resolution of 0.005°. The angle of the ankle was determined using a Delsys electrogoniometer (Boston, MA). One end of the electrogoniometer was securely fastened to the shank, while the other end was secured to the side of the foot. The sensors were then calibrated over a range of angles, using a protractor as an independent measure of angle. The sensitivity was approximately 50 °/V (depending on exact subject placement) and the precision was 0.5°.

B. Data Analysis

All data were low-pass filtered using a fourth order Butterworth filter with a cutoff frequency of 25 Hz, and segmented to include only the 150 ms of the ramp perturbation. The nominal forces—forces obtained while no subject was present—were subtracted from the force platform data for each trial, yielding the forces from the subject alone. Ankle response torque was determined using the inverted pendulum assumption [28], where the torque response is the product of the body weight and the change in the anterior-posterior position of the center of pressure (COP) (Fig. 3), determined relative to the initial position.

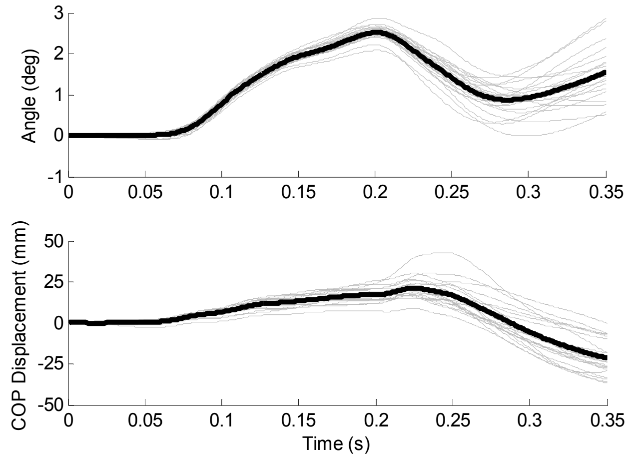

Fig. 3: Representative plots shown of a single subject's ankle angle and COP displacement used to determine ankle impedances. Individual trials are shown in grey and the average of all trials is shown in black. Note approximately 85% of the perturbation angle is absorbed by the ankle.

A second order parametric model was used to identify the impedance of the ankle

$$\delta T = I_{tot}\ddot{\theta} + b_a\dot{\theta} + k_a\theta\,, \qquad (1)$$

where δT is the torque response to the perturbation, I_{tot} is total inertia foot and other coupled body segments; and b_a and k_a are the damping and stiffness coefficients of impedance, respectively; finally, θ is the angular displacement of the ankle. The derivatives were computed using the bidirectional finite step method. The impedance parameters were identified by the least squares estimation method, over the 150 ms perturbation window. The impedance values were divided by two, to account for both feet. Variance accounted for (VAF) was used to quantify the agreement of the model with the experimental results.

IV. Results and Discussion

The impedance parameters obtained for each subject are shown in Table 1.

TABLE I: Average Results of Impedance Identification

Subject	I_{tot} (kg m^2)	b_a (Nm s/°)	k_a (Nm/°)	% VAF
S1	1E-4±0.01	0.03±0.03	4.3±0.9	98.4±1.7
S2	0.01±0.01	0.03 ±0.01	1.6 ±0.4	98.9±1.0
S3	0.02±0.01	0.02±0.01	2.6±0.6	98.3±2.0
S4	0.03±0.01	0.04±0.01	3.9±0.7	98.5±1.4

The subject's ankle damping coefficients ranged from 0.02 Nms/° to 0.04 Nms/° and their stiffness coefficients ranged from 1.6 Nm/° to 4.3 Nm/°. The inertial term has limited relevance due to its aggregate nature. The second

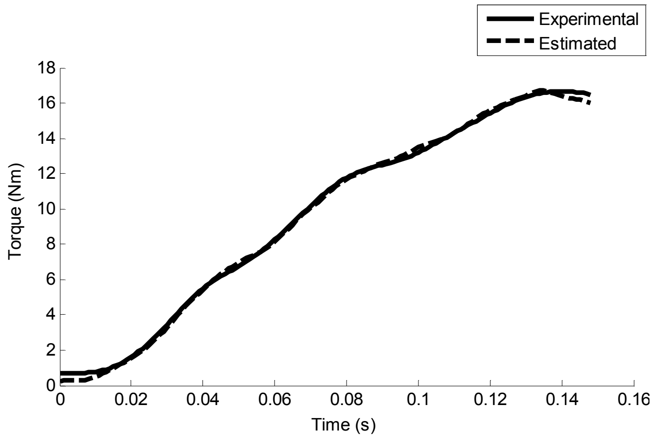

Fig. 4: Representative plot of a single trial showing the experimentally determined ankle response torque and the estimated ankle torque from the second order model. Note that the experimental torque was determined by the product of the subject's body weight and the anterior-posterior COP shift.

order model fit the data well with an average VAF of 98.5%. A representative plot is shown in Fig. 4.

In order to investigate the validity of the Perturberator, our estimates for ankle impedances were compared to those published in the literature (Table II).

TABLE II: COMPARISON WITH LITERATURE

Method	Study	Avg. k_a (Nm/°)
3° Dorsiflexive perturbation	Perturberator	3.1±1.2
0.055° Dorsiflexive perturbation	Loram & Lakie (2002)	5.2±1.2
1° Dorsi/plantarflexive perturbation	Casadio et al. (2005)	3.2±0.8

The results from the Perturberator compared favorably to those published in the literature. The ankle stiffness coefficient was less than the results published by Loram and Lakie [13]; however the perturbation implemented in that study is significantly smaller, and ankle stiffness estimates have been shown to increase with decreased perturbation amplitude [6]. Thus, with the agreement of the stiffness components of impedance, we believe the Perturberator to be a valuable measurement device that can accurately determine the impedance parameters of the ankle during standing.

This is a preliminary study and future work will focus on more subjects and validation with a mechanical system, further demonstrating the accuracy of the Perturberator. Subsequently, we will estimate the impedance parameters of the ankle during ambulatory tasks, including the stance phase walking, for the development of a biomimetic control system for powered prosthetic ankles.

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