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STOCHASTIC ESTIMATION OF HUMAN ANKLE MECHANICAL IMPEDANCE IN LATERAL/MEDIAL ROTATION

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ABSTRACT

This article compares stochastic estimates of human ankle mechanical impedance when ankle muscles were fully relaxed and co-contracting antagonistically. We employed Anklebot, a rehabilitation robot for the ankle to provide torque perturbations. Surface electromyography (EMG) was used to monitor muscle activation levels and these EMG signals were displayed to subjects who attempted to maintain them constant. Time histories of ankle torques and angles in the lateral/medial (LM) directions were recorded. The results also compared with the ankle impedance in inversion-eversion (IE) and dorsiflexion-plantarflexion (DP). Linear time-invariant transfer functions between the measured torques and angles were estimated for the Anklebot alone and when a human subject wore it; the difference between these functions provided an estimate of ankle mechanical impedance. High coherence was observed over a frequency range up to 30 Hz. The main effect of muscle activation was to increase the magnitude of ankle mechanical impedance in all degrees of freedom of ankle.

NOMENCLATURE

τ	Torque applied to the ankle
F_{right}	Force applied by the right actuator
F_{left}	Force applied by the left actuator
$D_{actuator}$	Distance between the 2 actuators
θ	Angle of the ankle
θ_{offset}	Initial offset of the ankle
X_{right}	Distance traveled by the right actuator
X_{left}	Distance traveled by the right actuator

INTRODUCTION

The ankle is of fundamental importance in different tasks such as locomotion and standing, given that it is the first joint to transfer the reactions forces from the ground to the rest of the body. Understanding the mechanical impedance of the human ankle may help to explain its role in the locomotion. During gait, the ankle rotates in all three anatomical planes, suggesting the mechanical impedance modulation occurs in all three degrees of freedom (DOF) of ankle. Ankle mechanical impedance has been studied in sagittal and frontal planes; however, there has been no reported estimation of ankle mechanical impedance in lateral/medial direction in transverse plane, which is presented in this paper.

Previous work estimated the multivariable mechanical impedance of ankle in IE and DP, when the muscles connected to the ankle joint were relaxed and in co-contraction [1-3]. Ankle impedance during muscle contraction is important since most activities are performed with varying levels of muscle activation. The muscle contraction can be monitored using Surface Electromyography (EMG). The sEMG signal amplitudes provide qualitative descriptions of the force and/or speed produced by the muscle and has a near monotonic relationship to muscle force in isometric contraction [4]. Stochastic identification method was used for estimation of multivariable mechanical impedance of ankle. The stochastic method does not require *a priori* assumption regarding the dynamic characteristics of the system, and so it is a suitable method to estimate the impedance of complex systems [5].

The mechanical characteristics of the ankle in the transverse plane are especially important during step turn since a torque is applied to the ankle to pivot the weight of the body when changing directions. Using a motion capture camera system, we determined the average ankle range of motion (Table 1) during straight steps and in sharp 90° contralateral step turns (Fig. 1) showing a decrease in the range of motion (ROM) in DP and ML during turning and an increase in range in IE. We speculated that a higher stiffness is required during turning to transfer the reaction torque from the ground to rotate the body; therefore, causing a smaller LM rotation compared to the straight walk. This was the motivation for estimating the ankle impedance in the transverse plane to incorporate the results in the design of a lower extremity prosthesis with appropriate lateral/medial ankle impedance.

TABLE 1. RANGE OF MOTION OF STRAIGHT WALK AND STEP TURN STANCE PERIODS

	ROM of Straight Step Stance Period (deg)		ROM of Step Turn Stance Period (deg)		% Change
	Degrees	Standard Error	Degrees	Standard Error	
DP	33.9	0.65	31.6	0.62	-7.4
IE	15.69	0.52	20.6	1.06	23.8
ML	22.09	0.6	16.8	0.65	-31.9

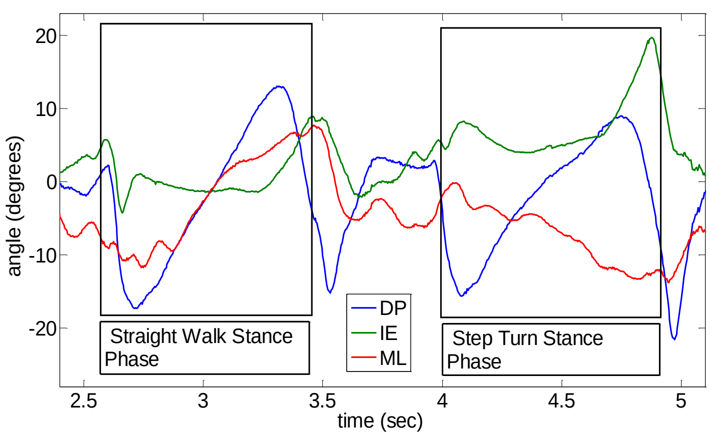

FIGURE 1. PLOT OF THE DP, IE, AND ML ROTATIONS OF A REPRESENTATIVE TEST DURING THE STREIGHT WALK STEP AND STEP TURN.

This paper reports the methodology and the preliminary results for estimation of the mechanical impedance of the ankle in the medial and lateral directions. A wearable ankle robot, Anklebot [5], was used to apply torque perturbations to the ankle and stochastic identification method was employed to estimate the ankle impedance with relaxed and active muscles in DP, ML, and IE.

EXPERIMENT METHODOLOGY AND RESULTS

Human Subjects

Five male subjects with no self-reported neuromuscular and biomechanical disorders were recruited for the experiments (ages from 23 to 26 years and body mass index from 18.5 to 27.5). The subjects gave written consents to participate in the experiment, which was approved by the Michigan Tech Institutional Review Board.

Experimental Setup

In previous work [1, 2], the Anklebot was used for applying the perturbation torques to the ankle in IE and DP directions simultaneously. For this experiment, the same procedures were applied to estimate the impedance in DP and IE. In addition, we modified the setup configuration and programmed the robot to generate torque perturbations only in the lateral/medial directions. The shoe and chair were modified to allow a horizontal placement of the Anklebot actuators (Fig. 3). Pseudo random voltage inputs with similar magnitude and opposite signs were applied to the actuators to generate a rotational motion of the foot about the tibia. The bandwidth of the input signals was 100 Hz which generated the medial/lateral rotation of the ankle with a root-mean-square (rms) of 5.5°, ensuring the induced rotations remain within the natural limits of the joint and preserve the linearity of the system. The shoe mounting bracket was fabricated and placed underneath the shoe. The robot was attached directly to a custom-made chair and the mounting bracket of the shoe while the participants were wearing it. The weight of the subject's leg and robot was supported by a knee brace and, to avoid forward and backwards swinging of the leg due to the actuator forces, a horizontal rod and a shin brace were used to stabilize the leg. Two spherical joints on each end of the rod assured no constraints in the medial/lateral rotation. The torque applied to the ankle and the angles of ankle were measured using the actuators' forces and displacements as follows:

$$\tau = \left(F_{right} - F_{left}\right) \times \frac{D_{actuator}}{2} \qquad (1)$$

$$\theta = \theta_{offset} + \arctan\left(\frac{X_{right} - X_{left}}{D_{actuator}}\right) \qquad (2)$$

where the parameters are defined in the nomenclature.

The muscles' activities were monitored using a Delsys ® Trigno wireless EMG system with surface electrodes placed at the bellies of the tibialis anterior (TA) and soleus (SOL) muscles. The EMG was sampled at 2 kHz and the rms value of a window of 13.5ms of data calculated and displayed as a visual feedback to the participant on a computer screen. Due to fatigue, a maximum voluntary contraction (MVC) experiment revealed 67% decay in the rms of the EMG from the TA muscle after a period of 60 seconds (Fig. 2); however, a 10% MVC of the TA was easily sustainable for the same period of time; therefore, the participants were instructed to maintain the 10% MVC of the TA during the experiments.

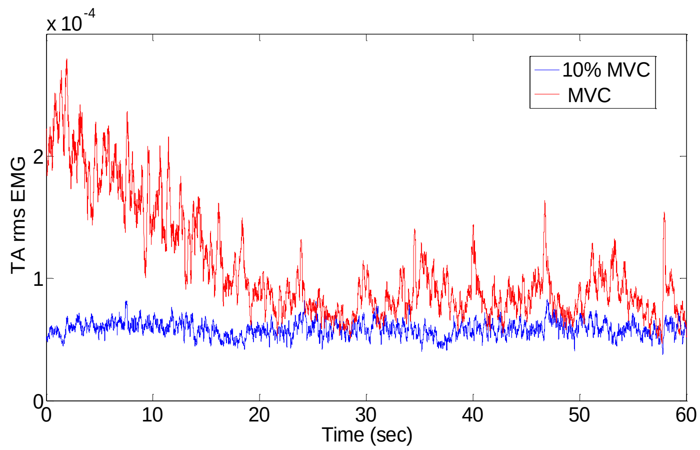

FIGURE 2. PLOTS OF TA EMG DECAY AT MVC AFTER A PERIOD OF 60 SECONDS OF A REPRESENTATIVE SUBJECT.

FIGURE 3. ANKLEBOT MOUNTED TO THE CHAIR FOR TESTING THE MECHANICAL IMPEDANCE IN THE MEDIAL/LATERAL ROTATION.

Experimental Protocol

To compare the results of the impedance in LM direction with the ankle impedance in DP and IE directions, we performed two separate experiments on the human subjects in LM and a multivariable estimation approach for DP and IE together as introduced in [1, 2].

Protocol 1: Impedance in LM Direction: Two different sets of experiments were performed to estimate the impedance in ML: one with all the leg muscles relaxed, and the other with TA and SOL co-contractions with 10% MVC. The impedance of the Anklebot was reduced from the impedance of the combined system of Ankle and Anklebot, as described in [1, 2].

Protocol 2: Impedance in DP/IE: In the DP/IE experiment the Anklebot was set to produce perturbations in both DP and IE directions. The protocol of the experiments was the same as previous work with the Anklebot [1, 2] with the perturbations applied for 70 seconds. The subjects were exposed to both relax and 10% TA co-contractions similar to the co-contraction ML tests.

Relax Muscle experiment. In the relaxed muscles study, the subjects were asked to maintain all their leg muscles relaxed. The EMGs of the SOL and TA muscles were visible to the participants at all times to assure no voluntary muscle contraction. The perturbations were applied for 70 seconds and the data from both the Anklebot and sEMG were recorded.

Co-Contraction Muscle experiment. Prior the active muscle study, the subjects were asked to produce the maximum voluntary muscle contraction for a period of 5 seconds to determine their EMG level during MVC. The impedance test was similar to the passive study except that the subjects were asked to co-contract the TA and SOL and keep 10% of the MVC of the TA. To help the subject to visualize the amount of the required muscle co-contraction a constant reference line equal to 10% of their TA MVC was plotted with their EMG signal.

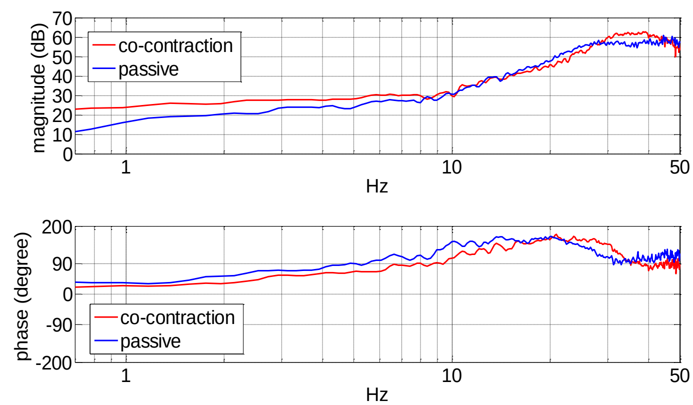

FIGURE 4. PLOTS OF THE MAGNITUDE AND PHASE OF THE IMPEDANCE IN MEDIAL/LATERAL ROTATION OF A REPRESENTATIVE SUBJECT.

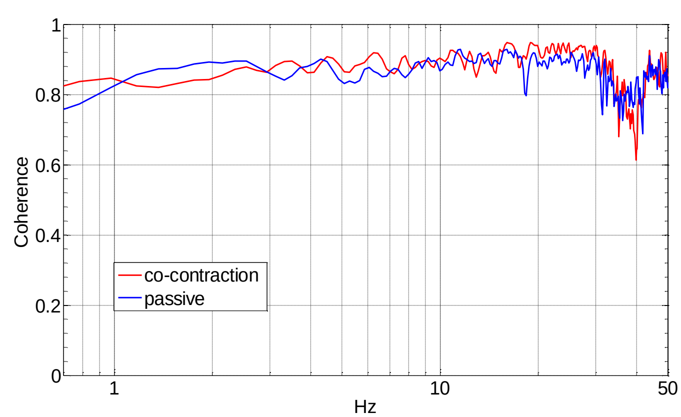

FIGURE 5. COHERANCE OF IMPEDANCE FOR BOTH RELAXED AND CO-CONTRATION TESTS OF A REPRESENTATIVE SUBJECT IN ML.

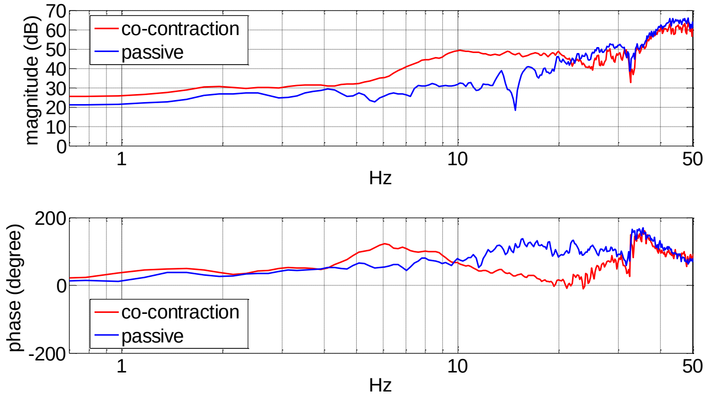

FIGURE 6. PLOTS OF THE MAGNITUDE AND PHASE OF THE IMPEDANCE IN DORSIFLEXSION/PLANTARFLEXION ROTATION OF A REPRESENTATIVE SUBJECT.

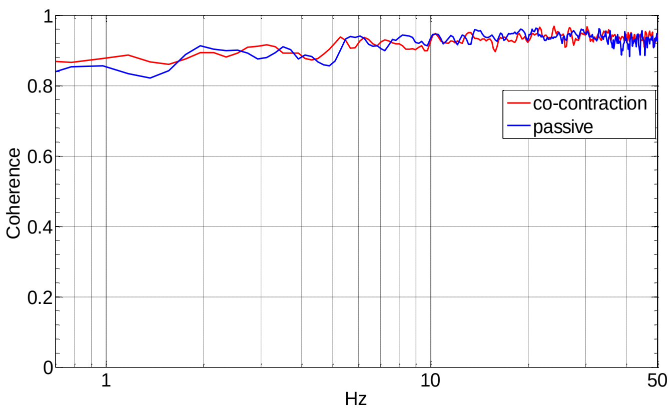

FIGURE 7. COHERANCE OF IMPEDANCE FOR BOTH RELAXED AND CO-CONTRATION TESTS OF A REPRESENTATIVE SUBJECT IN DP.

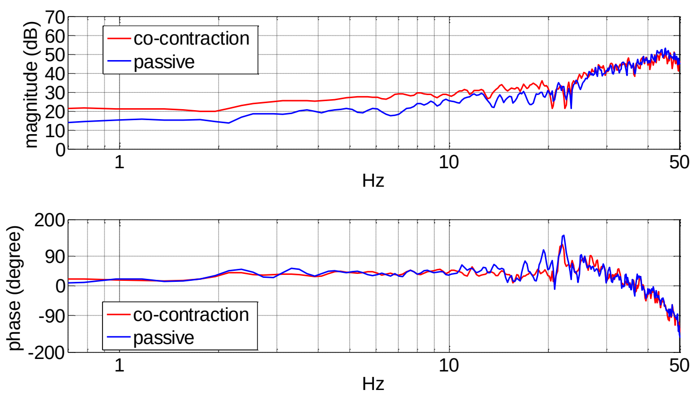

FIGURE 8. PLOTS OF THE MAGNITUDE AND PHASE OF THE IMPEDANCE IN INVERTION/EVERTION ROTATION OF A REPRESENTATIVE SUBJECT.

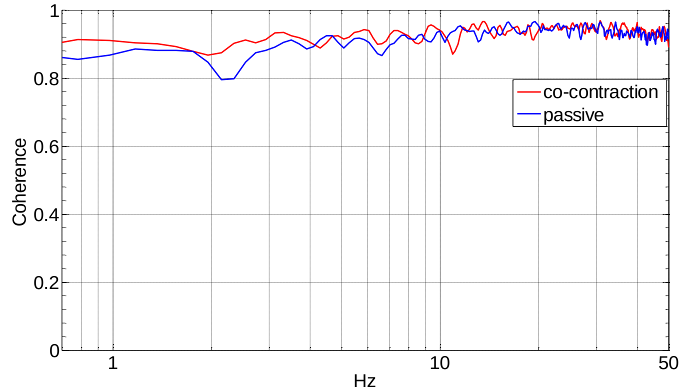

FIGURE 9. COHERANCE OF IMPEDANCE FOR BOTH RELAXED AND CO-CONTRATION TESTS OF A REPRESENTATIVE SUBJECT IN IE.

RESULTS AND DISCUSSION

Figure 4 shows the Bode plot of the ankle impedance in medial/lateral direction for a representative subject for both the active and passive tests. Figure 5 shows the coherences of the estimated frequency response. From the five subjects, the mean coherences were 0.87 and 0.88 in the frequency range of up to 50Hz for passive and active tests respectively. For up to 30 Hz, the mean coherences were 0.89 and 0.90 showing a dependable linear relation between torques and angles in both active and passive muscle tests within this frequency range.

The average break frequencies were 9.1Hz and 6.4 Hz for passive and active tests, respectively. This break frequency was consistent with the 90° phase crossing and showed similar behavior as IE and DP impedances [3]. On average, the impedance observed at frequencies below the break frequency for the contraction test was 96% higher with values of 28.6 dB (27 Nm/rad) and 22.8 dB (13.8 Nm/rad) for the active and passive tests, respectively. The slope of the magnitude plots in both relaxed and active experiments in the frequencies more than the associated break frequencies were 54.7dB/decade. These results are consistent with previous work in DP and IE where the viscoelastic characteristics of passive tissue and muscle activation direct the ankle behavior at low frequencies and foot inertia above the break frequency [3].The phase plot in Fig. 4 shows a consistent increase in phase for the passive impedance which is consistent with an expected lower response time with lower stiffness.

DP and IE tests showed similar changes in impedance as ML between relaxed and active tests (Fig. 6 and 8). DP impedance observed at frequencies below the break frequency for the co-contraction test were %86 higher with values of 31.2 dB (47.3 Nm/rad) and 26.9 dB (25.4Nm/rad) for the active and passive tests respectively. IE co-contraction showed an increase of %53 with values of 23.3 dB (18 Nm/rad) and 20.6 dB (11.8 Nm/rad) respectively. This indicates that ML has the highest increase in stiffness in co-contraction of the TA and SOL muscles and IE the least. Above break frequency, in both relax and muscle contraction, the slope of the magnitude in the DP was

43.587dB/decade and 44.2 dB/decade in IE. The impedance found in DP and IE both showed an average coherence of 0.92 (Fig. 7 and 9) and was not significantly affected by muscle contraction showing a dependable linear relation within this frequency range.

FUTURE WORK

Since it is of interest to develop prosthetic devices with anthropomorphic characteristics, passive or active elements can be incorporated in the design of powered prostheses to generate the desired amount of stiffness in the ankle joint. The impedance of ankle in this paper, however, was estimated during non-load bearing conditions. We will continue our study to estimate the time-vying impedance of ankle with similar estimation methods combined with an instrumented walkway as well as indirect methods utilizing a model of ankle impedance in terms of the statistical properties of EMG signals.

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