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STOCHASTIC ESTIMATION OF THE MULTI-VARIABLE MECHANICAL IMPEDANCE OF THE HUMAN ANKLE WITH ACTIVE MUSCLES

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ABSTRACT

This article compares stochastic estimates of multi-variable human ankle mechanical impedance when ankle muscles were fully relaxed, actively generating ankle torque or co-contracting antagonistically. We employed Anklebot, a rehabilitation robot for the ankle, to provide torque perturbations. Muscle activation levels were monitored electromyographically and these EMG signals were displayed to subjects who attempted to maintain them constant. Time histories of ankle torques and angles in the Dorsi-Plantar flexion (DP) and Inversion-Eversion (IE) directions were recorded. Linear time-invariant transfer functions between the measured torques and angles were estimated for the Anklebot alone and when it was worn by a human subject, the difference between these functions providing 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 both DP and IE directions.

INTRODUCTION

The mechanical impedance of the human ankle plays a major role in lower extremity functions that involve mechanical interaction of the foot with the contacted surface. Examples include maintaining upright posture and shock-absorption, lower-limb joint coordination, steering, and propulsion during walking on level ground, slopes or stairs. One method for measuring ankle mechanical impedance uses stochastic perturbations. It provides a quantitative estimate of ankle impedance without requiring a-priori assumptions about its dynamic structure. In particular, this method avoids the common assumption that impedance is composed of inertia, damping and stiffness, but is applicable to more complex, higher-order dynamics. A stochastic perturbation method was used by Kirsch et. al. [1] to estimate ankle mechanical impedance in one degree of freedom. Small stochastic motion perturbations were applied during a large dorsiflexion motion of the foot. Motion perturbations were used by Van der Helm et. al. [2] utilizing a linear hydraulic actuator to identify intrinsic and reflexive components of the human arm dynamics [2]. Applying motion perturbations requires care to avoid exerting excessive forces on subjects' joints. In earlier work, we employed MIT-MANUS to apply pseudo-random force perturbations to estimate the mechanical impedance of the arm in two degrees of freedom simultaneously [3]. We recently applied the same methodology to estimate the mechanical impedance of the relaxed human ankle in two degrees of freedom using Anklebot [4]. In this paper, we report application of the method to estimate the mechanical impedance of the human ankle with active muscles.

EXPERIMENTS

Human Subjects

Eight human subjects with no reported history of neuromuscular disorders (age range mid 20s to mid 30s) were recruited for this study. All subjects gave their informed consent prior to testing. The protocol was approved by the Massachusetts Institute of Technology Committee on the Use of Humans as Experimental Subjects (MIT-COUHES). Data from only one subject is presented in this preliminary report.

Experimental Procedures

The experimental procedure was similar to that reported in [4] to identify multi-variable ankle mechanical impedance in DP and IE simultaneously from nonparametric estimation of the best-fit linear transfer functions relating torques to angles in the DP and IE directions. We used Anklebot to apply perturbations and

measure ankle angles. Anklebot is backdrivable with low friction and allows human subjects to move their foot freely in three degrees of freedom (DOF) relative to the shank. Of those, two DOFs are actuated. Two nearly parallel actuators generate a dorsi-plantarflexion torque if both apply identical forces in the same direction, and inversion-eversion torque if they apply identical forces in opposite directions. As a result, the robot can apply simultaneous perturbations in two degrees of freedom of the ankle. Displacements of the linear actuators are measured by linear encoders. This information was recorded at a sampling rate of 200 Hz and the corresponding ankle torques and angles were estimated as described in detail in [4-5].

Ankle mechanical impedance was estimated in two steps: first, the impedance of the human ankle wearing the Anklebot was determined; second, the impedance of the Anklebot alone was subtracted from the first measurement. The result was the net impedance of the human ankle. During the first step, subjects wore Anklebot in a seated position. Two uncorrelated pseudo-random command voltages with bandwidth of 100Hz were applied to each Anklebot's actuators to produce torque perturbations. Torque perturbations moved a subject's foot in all directions in the frontal and sagittal planes while remaining within the natural limits of the joint. The magnitudes of the perturbations and evoked motions were small enough to ensure a linear relation between ankle torque and angle. A detailed description of the method can be found in Rastgaar et. al. [4].

Three electromyographic (EMG) sensors were attached to the subject's leg to monitor the activities of tibialis anterior, soleus, and peroneus longus. These muscles were selected based on their role in DP and IE movement of the foot. Tibialis anterior (TA) plays a major role in dorsiflexion and inversion. The main function of soleus (Sol) is plantarflexion. Peroneus longus (PL) is involved in plantarflexion and eversion. EMG signals were measured with a sampling rate of 200Hz. EMG amplitude was estimated using a root-mean-square filter with a window of 0.2 seconds. EMG amplitudes, as well as target bands denoting the limits of desired EMG variation, were displayed visually to subjects who were asked to maintain the desired muscle activity throughout the experiment. The EMG limits were chosen in such a way that maintaining the associated muscle activation was comfortable for the subject throughout the 60-second duration of the test. Five different tests were performed as follows: (1) fully-relaxed (passive) muscles; (2) active tibialis anterior; (3) active soleus; (4) active peroneus longus; and (5) co-contraction (with all muscles active). Table 1 shows the ratio of the RMS values of the EMG of the active muscles to the RMS value of the EMG of those muscles in passive test.

Analysis

Linear time-invariant transfer functions were estimated in the frequency domain in each direction by computing the ratio of the cross power spectral density of angle and torque to the power

Test \ Muscle	TA	PL	Sol
Active TA	12.5	1.1	0.6
Active PL	4.8	5.5	2.0
Active Sol	2.5	6.4	4.3
Co-contraction	8.0	2.3	3.8

TABLE.1. RATIO OF RMS EMG OF ACTIVE MUSCLES TO RELAXED MUSCLES

spectral density of angle [4].

RESULTS AND DISCUSSION

The goal of this study was to quantify and compare the multi-variable dynamic mechanical impedance of the ankle under different muscle activation conditions. Fig. 1 shows Bode plots of ankle mechanical impedance magnitude and phase in the DP direction for the five different muscle activation conditions for one subject. Fig. 2 shows the corresponding plots for the IE direction. Figs. 3 and 4 show partial coherence plots corresponding to Figs. 1 and 2 respectively.

The coherence of the linear transfer functions that describe the impedances in both the DP and IE directions were greater than 0.85 over the 0.5 to 30 Hz frequency range, confirming a reliable linear relation between torques and angles. Below 0.5 Hz, the coherences were relatively low, possibly due to nonlinearities of the electromechanical hardware (such as friction and motor cogging). Further investigation is ongoing.

The plots of ankle impedance in the DP direction with relaxed muscles showed that at frequencies above about 8 Hz, the magnitude plot rolled upwards with a slope approaching 40 db/decade (Fig. 1-A). The transition of phase angle from 0 to 180° passed through 90° at about 8 Hz (Fig. 1-B). This break frequency is consistent with foot inertia dominating ankle impedance above 8 Hz. Below this frequency, the viscous-elastic properties of passive ankle tissues and any active muscles play the dominant role. Plots of active-muscle ankle impedance in DP direction showed that break frequencies occurred at about 10 to 12 Hz. This is consistent with higher active-muscle ankle stiffnesses.

Ankle impedance in the IE direction (Figs. 2-A and 2-B) showed a consistently smaller magnitude than in the DP direction. Data above 10 Hz do not show a clear break point as in the DP direction and may reflect foot inertia or higher-order dynamics. Further investigation is ongoing.

Increasing muscle activity primarily increased the magnitude of ankle mechanical impedance. In DP, all active-muscle impedances exhibited a similar variation with frequency below about 10 Hz, which was different from the relaxed-muscle impedance. With active soleus or peroneus longus, DP impedance magnitudes were close to each other and, below about 10 Hz, both greater than the relaxed-muscle impedance.
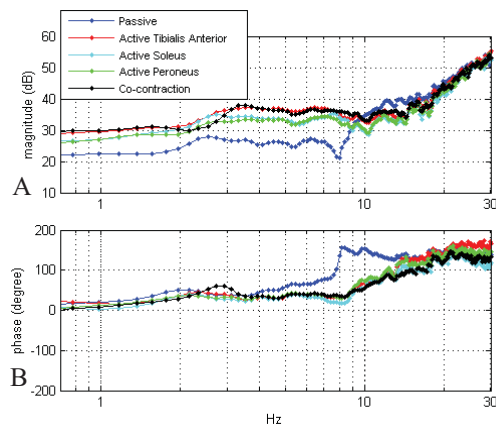
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FIGURE 1. A- MAGNITUDE AND B- PHASE PLOTS OF IMPEDANCES OF ANKLE IN DP DIRECTION WITH DIFFERENT ACTIVE MUSCLES

FIGURE 2. A- MAGNITUDE AND B- PHASE PLOTS OF IMPEDANCES OF ANKLE IN IE DIRECTION WITH DIFFERENT ACTIVE MUSCLES

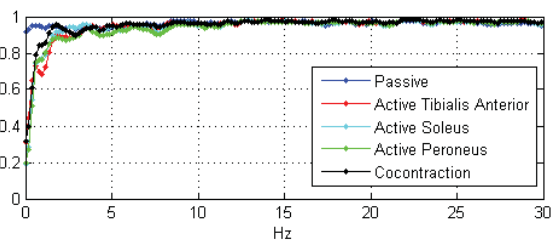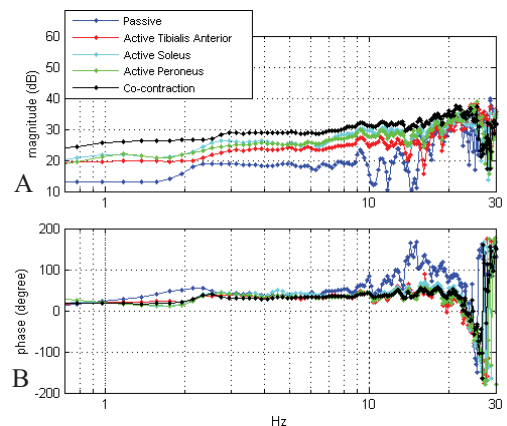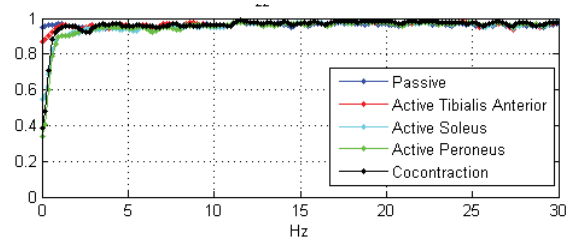FIGURE 3. COHERENCES OF IMPEDANCES IN DP DIRECTION WITH DIFFERENT ACTIVE MUSCLES

FIGURE 4. COHERENCES OF IMPEDANCES IN IE DIRECTION WITH DIFFERENT ACTIVE MUSCLES

Though both were greater than ankle impedance with relaxed muscles or with soleus or peroneus longus active, DP ankle impedance with active tibialis anterior was similar to ankle impedance with co-contraction of all three muscles. As shown in Table 1, the EMG level of tibialis anterior in the TA-active test, was greater than its level in the co-contraction test; however, the contributions of peroneus longus and soleus – plantar-flexor muscles – were also significantly increased in the co-contraction test. In IE, all impedances exhibited a similar variation with frequency below about 12 Hz. Interestingly, ankle impedance in the IE direction for the TA-active test was smaller than the other active-muscle tests, most likely because of the smaller contribution of peroneus longus that contributes to eversion. Further investigation is ongoing.

Taken together, these results confirm that multivariable stochastic estimation methods can reliably identify the effects of different active muscles on the mechanical impedance of the ankle over a relatively wide range of frequencies.

ACKNOWLEDGMENTS

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