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Static Ankle Impedance in Stroke and Multiple Sclerosis: A Feasibility Study

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Abstract—**Quantitative characterization of ankle mechanical impedance is critical for understanding lower extremity function in persons with neurological disorders. In this paper, we examine the feasibility of employing an ankle robot and multivariable analysis to determine static ankle impedance in 4 patients: 1 with multiple sclerosis and 3 with stroke. We employed a scalar based vector field approximation method which was successful in identifying young healthy subjects' ankle impedance. It enabled clear interpretation of spatial ankle impedance structure and intermuscular feedback at the ankle for both affected and unaffected legs. Measured impedance of two patients was comparable to healthy young subjects, while the other two patients had significantly different static ankle impedance properties.**

I. Introduction

The demand for lower extremity rehabilitation for people with neurological impairments is growing as the number of patients is increasing apace with the aging population. Every year, about 795,000 Americans have a stroke [1] and an estimated 800,000 Americans have cerebral palsy (CP) [2].

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In addition in the US, about 250,000~300,000 people have been diagnosed with multiple sclerosis (MS) [3], and 250,000 people have spinal cord injuries (SCI) [4]. Many of these persons have significant gait impairments including at the ankle, which plays a significant role during gait, contributing significantly to postural stabilization, propulsion, energy-absorption, and lower limb joint coordination during locomotion.

To understand the role of the ankle in lower extremity function, the static component of ankle mechanical impedance [5], a generalization of ankle stiffness, has been studied extensively for healthy [6-9] as well as neurologically impaired human subjects including stroke [10], CP [11], MS [12], and SCI [13] patients. Those studies considered only a single degree of freedom (DOF) in the sagittal plane (dorsiflexion-plantarflexion (DP)). Recently, we examined static ankle impedance for inversion-eversion (IE) and DP directions separately in subjects with chronic hemiparesis due to stroke [13]. To our knowledge, this is the first study characterizing ankle mechanical impedance in multiple DOFs on neurologically impaired subjects.

Previously we measured the static ankle impedance in young healthy subjects [5]. We formulated an impedance identification method as a vector field approximation problem, and enabled a new analysis of ankle impedance: identification of static impedance structure in coupled DOFs, precise quantification of the extent to which the ankle is "spring-like", and evidence of uniquely neural contribution.

Here we extend that work to neurologically impaired subjects. We investigated the impedance symmetry between both legs and examined the nature of inter-muscular feedback at the ankle by quantifying the curl component of the estimated vector field.

II. Methods

A. Subjects

Subject inclusion criteria for this study are as below.

1) Diagnosis of Parkinson's disease, MS, or stroke by a neurologist, or healthy individuals without other significant neurological problems.
2) Men and women between the ages of 18-89 years.
3) Must be able to walk 30 feet unassisted.

The participants in this study included 8 subjects with no history of neuromuscular disorders (4 males and 4 females;

age range mid 20's ~ mid 30's) and patients: 3 males and 1 female (Mean: 56.8 yrs.; SD: 10.2 yrs.). Three participants sustained a stroke (01, 02, 04) and one participant was diagnosed with MS (03). The Providence Veterans Affairs Medical Center (PVAMC) Institutional Review Board and MIT's Committee on the Use of Humans as Experimental Subjects approved the protocol, and informed consent was obtained for all participants.

B. Experimental Setup

We employed a wearable ankle robot, Anklebot (Interactive Motion Technologies, Watertown, MA, USA) as the centerpiece of our method of measuring multivariable static ankle impedance [14]. The Anklebot has two low-impedance active DOFs (sagittal and frontal motions) and a third passive DOF (transverse motion) which prevents imposing any inadvertent kinematic constraints on the motion of the ankle. This approach guarantees that no excessive forces are applied at the physical interface between human and machine. Here we employed a PD controller with proportional gain of 200 Nm/rad and derivative gain 1 Nm-s/rad. Subjects wore a modified shoe and a knee brace, to which the robot was connected, and then the knee brace was securely fastened to the chair to support the weight of the robot and the leg (Fig.1).

Fig. 1. Subject wearing Anklebot in a seated condition

Electromyographic (EMG) signals were recorded using surface electrodes with a bandpass between 20 and 450Hz (Delsys Inc, Boston, MA, USA) placed on four ankle movement prime mover muscles: tibialis anterior (TA), soleus (SOL), gastrocnemius (GAS), and peroneus longus (PL). EMG amplitude was calculated from raw data using a linear envelope method: full wave rectification followed by low-pass filter with cutoff frequency 10Hz.

C. Experimental Protocol

Subjects were seated with their ankles clear of the ground in an anatomically neutral position with the sole at a right angle to the tibia, and instructed to relax while the robot moved their ankles.

The protocol consisted of 24 movements along 12 equally-spaced directions in IE–DP space, with a nominal displacement amplitude of 20° in each direction. Perturbations began with 0° and ended at 330° at 30° increments. Note that 0° and 180° correspond to eversion and inversion, 90° and 270° correspond to dorsiflexion and plantarflexion, respectively. The Anklebot displaced the ankle with low speed (5°/sec), which was selected to avoid evoking stretch reflexes. Both legs were measured with 2 repetitions per side. Mechanical quantities were sampled at 200 Hz, and EMG data was recorded at 1kHz.

D. Analysis Method

Multivariable static ankle impedance was represented as a vector field (V) [5]:

$$(\tau_{IE}, \tau_{DP}) = V(\theta_{IE}, \theta_{DP}) \quad (1)$$

where θ_{IE} and θ_{DP} are angular displacements in the IE and DP directions, respectively, and τ_{IE} and τ_{DP} are the corresponding applied torques. In brief, the vector field approximation problem was decomposed into two scalar function (ϕ_1, ϕ_2) estimation problems (2) and each scalar function was identified by adopting the method of Thin-plate Spline (TPS) smoothing [16] with Generalized Cross Validation (GCV) [17].

$$\tau_{IE} = \phi_1(\theta_{IE}, \theta_{DP})$$

$$\tau_{DP} = \phi_2(\theta_{IE}, \theta_{DP}) \quad (2)$$

The approximated vector field was further decomposed into a conservative (zero curl) and a rotational field (zero divergence). The vector field approximation and decomposition methods are detailed elsewhere [5]. A single vector field was estimated by averaging 2 repeated measurements of each scalar component. For each of the 12 directions described earlier, displacements up to 20º were imposed and the corresponding torque field co-linear with the displacement was identified. The slope of the straight line that best fit this data in a least squares sense became the effective stiffness in that direction. Static Impedance symmetry between both legs was investigated. For each movement direction, the ratio of effective stiffness value of the more affected side to the less affected side was calculated and averaged across the 12 directions.

The neural contribution to ankle impedance can be studied from curl analysis of the decomposed vector field. If there is no intermuscular feedback between muscles, non-zero curl cannot be introduced either with intrinsic muscle mechanics or intramuscular feedback [15]. However, if intermuscular feedback exists and the contribution of feedback is not balanced, then non-zero curl components can be introduced. Thus, by investigating the non-zero curl components in the rotational field, we can assess the existence of unbalanced intermuscular feedback around the ankle. To investigate the non-zero curl components in the rotational field, the level of zero curl was defined by the vector field approximated from 5 repetitive measurements in load-free condition. The 95% confidence interval of the curl components of this field corresponds to -0.75 ~ 0.74 Nm/rad.

III. Results

The vector field was approximated based on friction-compensated measurements. The mean deviation between measurements and surface estimates was much smaller (< 0.01 Nm) than the apparatus measurement range (≈ 1 Nm [14]) due to the smoothing effect of the surface fitting procedure. A representative vector field having two scalar components and corresponding decomposed vector fields are presented in Fig.2.

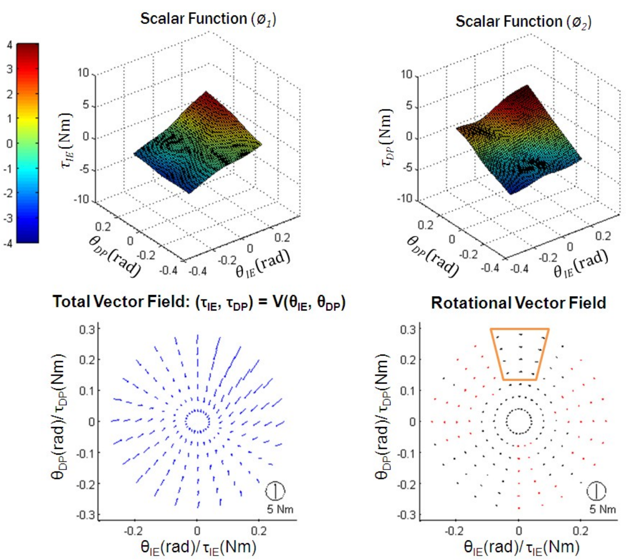

Fig. 2. A representative approximated vector field (consisting of two TPS surface estimates (ϕ_1, ϕ_2) based on 2 repetitive measurements (top) and the corresponding total and rotational field having non-zero curl components (bottom).

TABLE II
Rotational Field Analysis

Subject	Side	Mean Curl Outbound	Mean Curl Inbound	$\sqrt{K_{RATIO}}$
01	U*	-0.118 (0.359)	-0.205 (0.400)	0.436
	A*	-0.227 (0.337)	-0.184 (0.383)	0.151
02	U*	-0.254 (0.268)	-0.125 (0.375)	0.331
	A*	-0.233 (0.400)	-0.018 (0.437)	0.168
03	U	-0.086 (0.212)	-0.089 (0.259)	0.158
	A	-0.107 (0.189)	0.011 (0.251)	0.139
04	U	-0.128 (0.229)	-0.051 (0.184)	0.099
	A	-0.144 (0.162)	-0.073 (0.228)	0.098

A and U stand for affected and unaffected side, respectively. Mean of curl components in the rotational field was calculated for both outbound and inbound data. Outbound and inbound $\sqrt{K_{ratio}}$ values were averaged to a single value. Value in the parentheses is standard deviation. Result with * contains significant non-zero curl component.

The spatial ankle impedance structure of all subjects was illustrated for both legs in comparison with healthy young subject data from [5] (Fig.3), and the amount of symmetry between affected and unaffected sides was also quantified (see Table.1).

Subject 01 and 02 showed substantial asymmetry between more affected and less affected sides (greater than 2), with higher impedance values on the affected side. In contrast, subject 03 and 04 showed no significant difference (p >> 0.05).

Based on our criterion, non-zero curl components were detected in subject 01 and 02 for both more affected and less affected sides. The region in the box of rotational field in Fig.2 showed an example of significant non-zero curl components. The rotational fields of subject 03 and 04 were not significantly different from zero. To assess the size of non-zero curl component, the ratio between the square root of determinants of the anti-symmetric (rotational component) and symmetric parts (conservative component) of the stiffness matrix ($\sqrt{K_{ratio}} = \sqrt{\det(K_{anti-sym})}/\sqrt{\det(K_{sym})}$) was also calculated (Table. 2). $\sqrt{K_{ratio}}$ of the unaffected side for subject 01 and 02 was greater than 0.3.

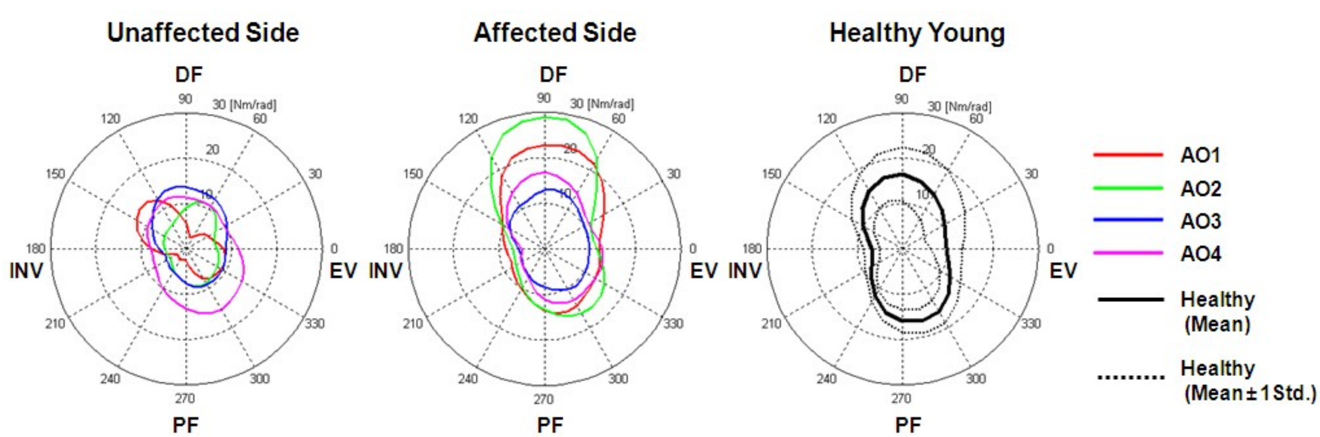

Fig. 3. Spatial ankle impedance structure of patients for both affected and unaffected sides and healthy young subjects in IE-DP space (EV: Eversion, INV: Inversion, DF: Dorsiflexion, PF: Plantarflexion). Outbound and inbound results were averaged.

IV. DISCUSSION

The feasibility of characterizing multivariable static ankle impedance for neurologically impaired subjects was demonstrated in this pilot experiment with 4 patients (3 stroke and 1 MS). We employed the same scalar based vector field approximation, which was successfully employed with young healthy subjects. It allows us to assess the amount of symmetry and the probe for the existence of abnormal intermuscular feedback at the ankle.

Static ankle impedance varied substantially with direction, exhibiting a characteristic "peanut" shape, pinched in the direction close to the IE axis. The structure of subject 03 and 04 was comparable with the results of young healthy subjects for both legs. However, results of subject 01 and 02 were significantly different from those of young healthy subjects. Impedance values on the more affected side, especially in the dorsiflexion direction, were substantially higher. These results are consistent with the previous results on stroke survivors with age-matched controls [18]. Interestingly, the impedance shape of subject 01 on the non-paretic ankle was tilted substantially (about 45°) in a counterclockwise direction. This might be a result of an abnormal gait pattern.

Non-zero curl components were detected in subjects 01 and 02 for both affected and unaffected sides. This is in contrast to results obtained in young healthy subjects that were conservative everywhere. If the damage to descending neural pathways, such as may occur in stroke, affects peripheral neural networks and causes unbalanced intermuscular feedback accordingly, that could account for these non-zero curls. The size of non-zero curl components was assessed using $\sqrt{K_{ratio}}$ index. The values for subjects 01 and 02 were higher than those for subjects 03 and 04 due to higher curl values in the rotational field in the subjects 01 and 02. Significant differences between affected and unaffected sides in subject 01 and 02 may be attributed to the larger conservative components because of increased impedance on the affected side.

Another interesting observation was that non-zero curl components were detected only in the case where the ankle impedance structure was asymmetric. Further study is needed to verify any correlation between curl components and the amount of symmetry of the impedance structure.

Why are the asymmetry of impedance structure and non-zero curl components important? Asymmetric ankle impedance may require a more complex control scheme to guarantee balance, postural stabilization and joint coordination during locomotion. Quantifying the curl components is also important because they are non-passive, and may function as an energy source or a sink under certain conditions, which may affect the stability of motor behavior. Thus, we anticipate that a quantitative characterization of multivariable static ankle impedance will provide a way to understand lower extremity functions of neurologically impaired patients, and eventually to explore strategies to rehabilitate them. In addition, we plan to study multivariable dynamic ankle impedance of neurologically impaired subjects.

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Clinical testing and evaluation of participants was conducted at the Neurorehabilitation Laboratory, Providence VA Medical Center.

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