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QUANTITATIVE CHARACTERIZATION OF STEADY-STATE ANKLE IMPEDANCE WITH MUSCLE ACTIVATION

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ABSTRACT

Characterization of multi-variable ankle mechanical impedance is crucial to understanding how the ankle supports lower-extremity function during interaction with the environment. This paper reports quantification of steady-state ankle impedance when muscles were active. Vector field approximation of repetitive measurements of the torque-angle relation in two degrees of freedom (inversion/eversion and dorsiflexion/plantarflexion) enabled assessment of spring-like and non-spring-like components. Experimental results of eight human subjects showed direction-dependent ankle impedance with greater magnitude than when muscles were relaxed. In addition, vector field analysis demonstrated a non-spring-like behavior when muscles were active, although this phenomenon was subtle in the unimpaired young subjects we studied.

INTRODUCTION

Ankle mechanical impedance, which we define as a functional that maps a time history of angular motion of the ankle joint onto a corresponding ankle torque time-history, plays a significant role in natural interaction of the lower extremity with the environment, including postural stabilization during standing and propulsion, energy-absorption, and lower-limb joint coordination during locomotion.

Most prior studies of human ankle mechanical impedance focused on the sagittal plane (the dorsiflexion/plantarflexion (DP) direction) [1−3]. To the best of our knowledge, only two studies have measured ankle impedance in the frontal plane (the inversion-eversion (IE) direction) [4−5]. Considering that single degree of freedom (DOF) movements at the ankle are rare under natural physiological conditions, characterization of ankle impedance in multiple DOFs promises deeper understanding of its roles in lower extremity function. In a previous work by the authors [6], the multivariable steady-state torque-angle relation at the ankle was measured with muscles maximally relaxed, showing that (as expected) the ankle behaved like a (nonlinear) spring under those conditions. Although that work provided a baseline for understanding ankle impedance, it is not directly applicable to normal lower-extremity actions since they involve muscle activation, either singly, synergistically or antagonistically (co-contraction). Here we extend the characterization of steady-state multi-variable ankle mechanical impedance to muscle active conditions.

EXPERIMENTS AND ANALYSIS METHODS

Human Subjects

Eight human subjects with no history of neuromuscular disorders (4 males and 4 females; age range mid 20's ~ mid 30's) were recruited for this study. Participants gave informed consent as approved by MIT's Committee on the Use of Humans as Experimental Subjects (COUHES).

Experimental Setup

Steady-state torque-angle data in IE−DP space were captured using a wearable ankle robot, Anklebot [7], the same device used in previous work [6]. It was mounted at the knee and operated by a simple impedance controller which enabled stable data capture even in high muscle activation conditions.

Surface electromyographic (EMG) signals were recorded from four major muscles related to ankle movement: tibialis anterior (TA), soleus (SOL), gastrocnemius (GAS), and peroneus longus (PL). EMG amplitude was calculated from raw data (sampled at 200Hz) using a root-mean-square filter with a window of 0.2 seconds.

Experimental Protocol: Active Study

Subjects were seated and asked to activate a specific muscle and maintain it at a constant level as best they could. The current EMG amplitude and target bands representing desired EMG amplitude were displayed in real time using an oscilloscope. To minimize effects due to inconstant muscle activation during the protocol, 5 repetitive measurements were performed for each active study. There were 3 types of active study: 2 single muscle activations (TA active and SOL active) and 1 co-contraction. Eight directions in the IE−DP plane (eversion, eversion + dorsiflexion, dorsiflexion, dorsiflexion + inversion, inversion, inversion + plantarflexion, plantarflexion, plantarflexion + eversion) were selected for the measurement, and terminated ramp perturbations were applied along these directions. Displacement amplitude was selected as 20° to cover the normal range of motion of the ankle. To maintain quasi-static conditions and avoid evoking stretch reflexes, movement velocity was regulated not to exceed 5°/sec. During perturbations the angular displacement and applied torque in both DOF as well as EMG data were recorded at 200 Hz.

Vector Field Approximation and Decomposition

Steady-state ankle impedance at the ankle was represented as a vector field (V), and this was approximated based on the methods introduced in our previous work [6]. In brief, the vector field approximation problem was decomposed into two scalar function ($\emptyset_1, \emptyset_2$) estimation problems (Fig.1) and each scalar function was identified by adopting the method of Thin-plate Spline (TPS) smoothing with Generalized Cross Validation (GCV). 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 in [6].

$$\boxed{(\tau_{IE}, \tau_{DP}) = V(\theta_{IE}, \theta_{DP})} \Rightarrow \boxed{\begin{aligned} \tau_{IE} &= \phi_1(\theta_{IE}, \theta_{DP}) \\ \tau_{DP} &= \phi_2(\theta_{IE}, \theta_{DP}) \end{aligned}}$$

Fig.1. Vector field approximation based on scalar methods (θ_{IE} and θ_{DP} are angular displacement in IE and DP direction, respectively, and θ_{IE} and θ_{DP} are applied torque at θ_{IE} and θ_{DP})

A single vector field was estimated by averaging 5 repeated measurements of each scalar component. Ankle impedance quantification and further analysis of the vector field were based on these averages.

What Can We Learn From the Vector Field?

First, spatial ankle impedance structure can be identified from the approximated vector field, since torque values (τ_{IE}, τ_{DP}) at any position (θ_{IE}, θ_{DP}) in the IE-DP space are easily calculated, which means the steady-state ankle impedance can be evaluated over the full range of motion, including 8 directions. The ankle impedance estimate is robust even with noisy data, since our methods are based on average and smoothed scalar functions.

Decomposed vector fields provide a precise quantification of the extent to which the ankle is spring-like. Specifically, if the set of active muscles behaves like a spring, the vector torque field must be an integrable function of displacement, the gradient of a potential function and, as a result, the curl of the torque field must be identically zero. Thus we can quantify the spring-like property of the ankle by comparing the size of the rotational field relative to the conservative field.

We can also investigate the role of intermuscular feedback from the rotational field. If there is no intermuscular feedback between muscles, that is the action of one muscle group does not influence the action of others, non-zero curl cannot be introduced either with intrinsic muscle mechanics or intramuscular feedback. However, if intermuscular feedback exists and the contribution of feedback is not balanced, then non-zero curl components can be introduced ($\partial F_i/\partial q_j \neq \partial F_j/\partial q_i$). Thus, by investigating the non-zero curl components in the rotational field, we can assess the existence of unbalanced intermuscular feedback around the ankle.

RESULTS AND DISCUSSION

To check the validity of these active studies, the ratio of active EMG levels to passive levels was calculated. The average result for the 8 subjects shows that subjects could follow instructions well (Table. 1).

Study \ Muscle	TA	PL	SOL	GAS
TA Active	**6.41(0.90)**	2.56(1.05)	1.02(0.08)	0.98(0.11)
SOL Active	1.38(0.10)	3.14(1.29)	**2.95(0.30)**	2.02(0.38)
Cocontraction	**9.77(1.43)**	**4.56(1.35)**	**3.52(0.36)**	**2.43(0.56)**

Table.1. Ratio of active EMG levels to passive levels (Shaded: muscles asked to activate, ( ): standard error)

Analysis of variance (1-way ANOVA) was performed to check the variation of EMG level in the 5 repetitive measurements both within and between subjects. In general, subjects could maintain constant muscle activation level across 5 repetitive measurements. Within subject analysis, all subjects showed no statistical difference in TA activation levels (p-value > 0.05). 7 and 5 subjects showed no statistical difference in SOL active and cocontraction case, respectively. However, there exists significant difference (p-value << 0.05) in the mean level of muscle activation between subjects in all 3 types of muscle activation. This is because the preferred activation level was selected by subject around the given reference level not exceeding the limit of the robot.

The spatial impedance structure at the ankle was constructed by calculating impedance magnitude (a 1-DOF linear approximation of ankle stiffness) in each of 24 directions from the vector field. The averaged results from 8 subjects in the 3 active studies are presented in Fig.2.

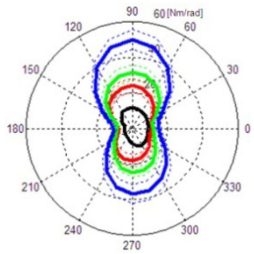

Fig.2. Spatial ankle impedance structure
(Solid line: averaged value, Dashed line: average±standard error, Black: passive, Red: TA active, Green: SOL active, Blue: Cocontraction)

Both single muscle activation and co-contraction increased ankle impedance in all directions by a factor of 2 and 3 (respectively) over the maximally-relaxed results. We can also see that impedance increases less in the frontal plane than in the sagittal plane in all active cases. As a result the pinched "peanut-shaped" structure is evident and even enhanced in all three muscle-active conditions. This intriguing result may account for the prevalence of ankle injury in the IE direction (rather than the DP direction) since the small impedance may be insufficient to stabilize the joint against external perturbations.

Fig.3 shows the results of vector field decomposition for a representative subject.

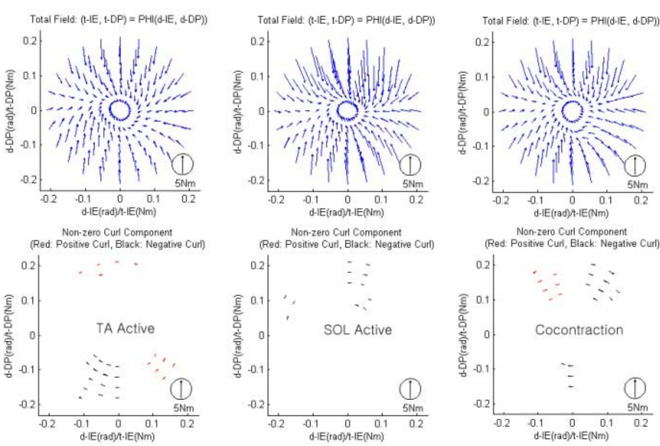

Fig.3. Vector field analysis of active studies (1st, 2nd and 3rd column: TA active, SOL active, Cocontraction, 1st and 2nd row: total field, rotational field)

Contrary to the maximally-relaxed result reported in [6], significant non-zero curl components were detected in the rotational field when muscles were active. This was true for all 8 subjects and for all 3 types of active study: muscle activation evoked significant non-spring-like behavior at the ankle. The non-zero rotational field (non-zero curl components) supports the existence of unbalanced intermuscular feedback at the ankle when muscles are active. However, we found no common patterns in the rotational field across subjects, suggesting that this observation may be due to imperfect tuning of spinal feedback circuits (perhaps due to the unfamiliarity if the task). How important is non-spring-like behavior? To assess the importance of non-spring-like behavior, the ratio between the determinants of the anti-symmetric (rotational component) and symmetric parts (conservative component) of the stiffness matrix was calculated. The result showed that rotational components are less than 10% of conservative components, meaning that any non-spring-like behavior of the ankle is subtle in the unimpaired young subjects we studied. From an engineering point of view we can also interpret this result to mean that (for healthy young subjects) ankle impedance with muscles active can be modeled as a nonlinear spring with less than 10% error.

FUTURE WORK

We continue to investigate the relation between muscle activation level and ankle impedance. As future directions, we plan to study the ankle impedance of elderly and neurologically impaired subjects. In addition, we are developing analysis methods and experimental protocols to explore transient dynamic ankle impedance which can be applied to general lower extremity functions.

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