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Relationship between Ankle Stiffness Structure and Muscle Activation

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Abstract—**This paper presents a characterization of the structure of ankle stiffness under multiple levels of muscle activation and the relationship between them. A multi-variable impedance estimation method using a wearable ankle robot enabled clear identification of ankle stiffness structure in the space consisting of the sagittal and frontal planes. With visual feedback showing current and target muscle activation levels, all subjects could successfully maintain multiple target levels (5%~30% of the maximum voluntary contraction level). Stiffness increased with muscle activation, but the increase was more pronounced in the dorsiflexion-plantarflexion direction than in the inversion-eversion direction, which resulted in a characteristic "peanut" shape. The relation between measured muscle activation level and ankle stiffness was evaluated. All subjects showed a highly linear relation not only for the two principal axis directions of the ankle, i.e., dorsiflexion-plantarflexion and inversion-eversion, but also for the average stiffness value of all directions. These major findings were consistent both for the tibialis anterior and triceps surae activation.**

I. Introduction

Mechanical impedance of the human ankle has been studied extensively for its importance in the natural interaction of the lower extremities with the environment. To better understand how ankle impedance changes with muscle activation, it has been investigated in various measurement conditions: seated [1, 2], supine [3, 4], quiet standing [5, 6], and running [7, 8].

While most of the previous studies have focused only on a single degree-of-freedom (DOF) of the ankle, especially in the sagittal plane, the authors' group recently has examined multi-variable ankle impedance in coupled DOFs, i.e., combinations of the sagittal and frontal plane motions [2, 9, 10]. Investigation of multiple DOF properties is important because the ankle is a biomechanically complex joint [11] and normal lower extremity actions, such as walking, involve substantial multiple DOF movements of the ankle, requiring coupling of dorsiflexion-plantarflexion (DP) and inversion-eversion (IE) [12].

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In our previous work, we identified the static component of ankle impedance, and characterized ankle stiffness structures of young healthy human subjects under fully relaxed [9] and active muscle conditions [2]. In all measurement conditions, we found an interesting stiffness structure with a characteristic "peanut" shape, being weakest in inversion direction. However, in our earlier active muscle study, only a single low target activation level (10% of the maximum voluntary contraction (MVC) level) was used, which was not enough to investigate the effect of different muscle activation levels on ankle stiffness structure. In some studies of single DOF movements it has been shown that ankle stiffness in the sagittal plane is proportional to applied ankle torque [3, 4], but the relationship in multiple DOFs has not been explored.

In this paper, we extend our previous work to measurements under multiple levels of muscle activation (5%~30% of the MVC level), and we examine the relationship between muscle activation and ankle stiffness structure in the sagittal and frontal planes.

II. Methods

A. Subjects

The participants in this study included 11 subjects with no history of neuromuscular disorders (6 males and 5 females; age range early 20's ~ early 30's). 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

A wearable robot, Anklebot, and electromyographic (EMG) sensors were used to investigate the relationship between muscle activation and ankle stiffness structure. The Anklebot (Interactive Motion Technologies, Watertown, MA, USA), which applies torque perturbations to the ankle in both DP and IE directions, enables reliable estimation of the stiffness structure. The robot was mounted to the knee brace, and two linear actuators were connected to the custom shoe. Measurements were performed in a seated posture while the weight of the robot was supported through connection to the side plate of the chair (Fig.1).

The Myomonitor wireless EMG system (Delsys, Boston, MA, USA) was used to record muscle activation levels during measurements. Surface electrodes with bandwidth ranging from 20 to 450 Hz and 16-bit accuracy were attached to four primary muscles involved in ankle movements: Tibialis

Fig.1. A subject wearing the Anklebot in a seated posture (left: front view, right: side view)

Anterior (TA), Soleus (SOL), Gastrocnemius (GAS), and Peroneus Longus (PL).

Both torques and angular displacements in 2 DOFs (recorded from Anklebot sensors) and EMG signals were sampled at 1 kHz. EMG amplitudes were calculated from the sampled raw signal using a root-mean-square filter with a moving window of 500 ms.

C. Experimental Protocol

Subjects were asked to sit with their ankle clear of the ground in a neutral position that the tibia was perpendicular to the sole. To select target muscle activation levels, the MVC level of each muscle was measured first. Subjects were asked to activate the muscle to their maximum level and maintain it for 5 seconds while the robot held the ankle near the neutral position. To provide enough restoring torque, the Anklebot stiffness was set to 2000 N/m for each actuator. Measurements were repeated 3 times for each muscle with enough rest time between measurements to minimize fatigue. The MVC level was determined as the mean of 3 measurements.

As a baseline for active studies, ankle impedance was first measured with fully relaxed muscles. A single measurement lasted for 40 seconds while the robot applied random torque perturbations (with bandwidth of 100 Hz) to the ankle in both DOFs.

Then, subjects were trained and instructed to activate a specific muscle and maintain it at a target activation level as best they could. To investigate effects of muscle activation on ankle stiffness, ankle impedance was measured at 6 different activation levels (5% to 30% of MVC level in increments of 5% MVC). A visual display showing current and target activation levels were provided to subjects. Both a dorsiflexor and a plantarflexor were studied, and the TA and the triceps surae (TS) were selected as target muscles for dorsiflexion and plantarflexion, respectively. For the TS study, the SOL was initially targeted but when a subject failed to maintain a constant activation level of the SOL, the GAS muscle was selected instead. To prevent muscle fatigue, a 3 minute rest period was given to each subject between measurements of 5% and 20% MVC, and 5 minute rest was given between intervals 20% and 30% MVC.

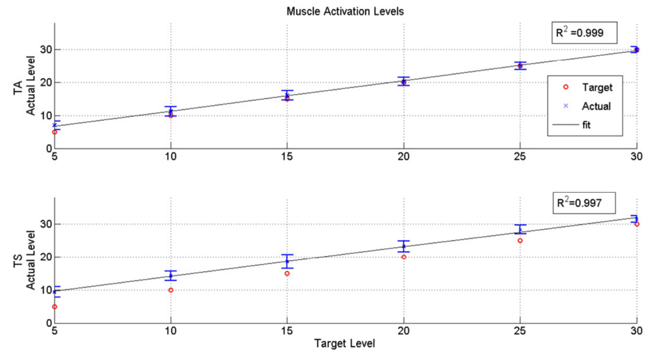

Fig.2. Actual muscle activation levels vs. Target levels in % MVC (Top: TA active study, Bottom: TS active study). Target levels are represented as red circles. Mean and standard deviation (SD) of all subjects are illustrated as blue cross marks and bars, respectively. Solid black lines show a linear fit of actual muscle activation levels vs. % MVC. The correlation coefficient (R^2) for the linear fitting was provided.

D. Analysis Methods

Multi-variable ankle mechanical impedance can be identified from the time history of angular displacement $(\boldsymbol{\theta} = (\theta_{DP}, \theta_{IE}))$ and torque data $(\boldsymbol{\tau} = (\tau_{DP}, \tau_{IE}))$ using a stochastic identification method [13, 14]. The contribution of the actuator dynamics was compensated by subtracting an impedance model of the actuator, which was obtained by running the same procedure but with no connection to a human subject, from the measured impedance. Details of identification methods are fully described in [15, 16].

Ankle impedance in any direction of coupled DOFs can also be identified by simple rotation operations. With a rotational transformation $(\boldsymbol{R})$, the new coordinate $(\boldsymbol{\theta}')$ and the corresponding torque $(\boldsymbol{\tau}')$ were defined (Eq.(1)).

$$\boldsymbol{\theta}' = \boldsymbol{R}\boldsymbol{\theta},\ \boldsymbol{\tau}' = \boldsymbol{R}\boldsymbol{\tau},\ \boldsymbol{R} = \begin{bmatrix} \cos\alpha & -\sin\alpha \\ \sin\alpha & \cos\alpha \end{bmatrix} \tag{1}$$

$$\boldsymbol{\theta}' = (\theta'_{DP}, \theta'_{IE}),\ \boldsymbol{\tau}' = (\tau'_{DP}, \tau'_{IE})$$

where α is the angle defined as a counter clockwise direction from the axis for the direction of θ_{DP}. By changing α from 0° to 90° incrementally (10°) and applying the impedance identification method to the transformed data ($\boldsymbol{\theta}'$ and $\boldsymbol{\tau}'$), we can identify ankle impedance in any direction in the 2D-space formed by θ_{DP} and θ_{IE} axes. This study focused on the static component of ankle impedance (stiffness[1]),

[1]Strictly speaking, stiffness and static mechanical impedance are different. Stiffness is a linear approximation to static mechanical impedance. But in this study, we assumed the term “ankle stiffness” means static component of ankle mechanical impedance.

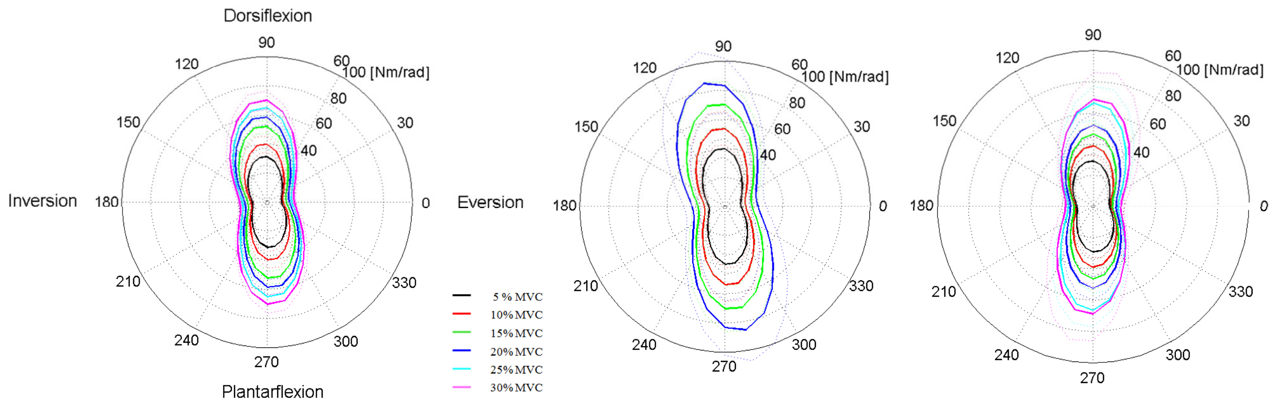

Fig.3. Variation of ankle stiffness structure with muscle activation. Left: TA active study, Middle: TS active study (1st set), Right: TS active study (2nd set). The angle of the polar plot represents movement direction of the ankle: 0°, 90°, 180° and 270° correspond to the eversion, dorsiflexion, inversion and plantarflexion direction, respectively. The radius of the plot depicts the stiffness value. The solid line is the mean value for all analyzed subjects, the dashed line denotes mean±standard error.

TABLE I
RATIO OF ANKLE STIFFNESS INCREASE WITH MUSCLE ACTIVATION

Target Activation Level	TA Study			TS Study (All subjects)			TS Study (4 subjects)		
	DP	IE	All	DP	IE	All	DP	IE	All
5	1.51	1.11	1.26	2.02	1.30	1.55	1.36	1.15	1.23
10	1.96	1.26	1.51	2.46	1.44	1.80	1.82	1.27	1.47
15	2.62	1.52	1.91	3.27	1.99	2.43	2.18	1.35	1.66
20	2.92	1.73	2.14	4.07	2.67	3.15	2.45	1.59	1.91
25	3.27	1.93	2.38	-	-	-	3.12	1.79	2.28
30	3.51	2.13	2.6	-	-	-	3.24	2.00	2.46

Ratio of ankle stiffness in active conditions to the relaxed condition was evaluated for each movement direction of coupled DOFs, and results on principal directions (DP and IE), and average of all directions (All) are summarized.

TABLE II
CORRELATION COEFFICIENT (R^2) OF THE LINEAR FIT

Subject	TA Study			TS Study		
	DP	IE	All	DP	IE	All
1*	0.95	0.94	0.95	0.84	0.92	0.88
2	0.96	0.87	0.97	0.99	0.72	0.97
3*	0.89	0.91	0.91	0.71	0.86	0.79
4	0.95	0.93	0.99	0.90	0.94	0.92
5	0.91	0.86	0.95	0.92	0.69	0.94
6	0.93	0.92	0.93	0.93	0.88	0.90
7	0.98	0.97	0.99	0.71	0.97	0.96
8*	0.96	0.93	0.96	0.98	0.65	0.92
9	0.82	0.95	0.89	0.89	0.92	0.80
10*	1.00	0.93	1.00	0.98	0.96	0.97
Mean	**0.93**	**0.92**	**0.95**	**0.88**	**0.85**	**0.90**
SD	**0.05**	**0.03**	**0.03**	**0.10**	**0.12**	**0.06**

All activation level (5 to 30% MVC) data were used in the analysis for the asterisk (*) denoted subject. Unmarked subjects used levels of 5 to 20% MVC data.

which we defined as the average impedance below 3 Hz.

The relation between muscle activation levels and ankle stiffness was evaluated for the principal directions (DP and IE) and the average of all directions.

III. RESULTS

A. Muscle Activation Levels

To ensure that each subject was able to maintain target muscle activation levels, the mean EMG amplitude for each measurement was calculated and its ratio to the MVC level was compared with the corresponding target level. One subject (#7), who showed abnormal EMG variability, was excluded from the data analysis. All subjects had no problem in activating the TA muscle, but 3 subjects showed difficulties in controlling the SOL. Therefore, for the TS active study, these 3 subjects targeted the GAS instead.

All subjects were able to maintain linearly increasing TA target levels (Fig.2). Activation levels of TS were slightly higher than the target levels, but still increased linearly with % MVC (R^2=0.997).

B. Ankle Stiffness Structure vs. Muscle Activation Levels

For each measurement condition, ankle impedance identifications were performed for different directions of the coupled 2 DOFs ($\alpha = 10°, 20°, \ldots, 90°$) , and the corresponding stiffness values were calculated. For the TS active study at 25% and 30% of MVC, the ankle torque of many subjects (7 out of 11) reached the actuator torque limit; hence these identification results should be interpreted with caution. Two sets of results were presented for the TS active study: the first set contained all subjects' muscle activation levels ranging from 5% to 20% of MVC; the second set contained 4 subjects' activation levels ranging from 5% to 30 % of MVC.

The ankle stiffness structures were plotted in polar coordinates (Fig.3). For both TA and TS active studies, the ankle impedance structure expanded in all directions with increasing muscle activation. In addition, larger expansion of the stiffness structure was found in DP than IE (Table I).
Linearity between measured muscle activation level and ankle stiffness was evaluated by calculating the Pearson correlation coefficient (R^2). All subjects showed a clear linear

relationship (high R^2 value close to 1) for not only the DP and IE directions but the average of all directions (Table II).

IV. Discussion

An accurate characterization of ankle stiffness structure with active muscles is important to better understand the role of the ankle in its interaction with the environment, since normal lower extremity functions involve ankle movements in coupled DOFs with different levels of muscle activation.

Use of a wearable robot, which actuates the ankle in 2 DOFs (both DP and IE directions), with a multi-variable stochastic identification method enabled clear identification of ankle stiffness structures under different levels of muscle activation. To our knowledge, this study is the first trial to examine the ankle stiffness structure in multiple DOFs with various muscle activation levels (5% to 30% of the MVC level).

With the suggested experimental setup and procedure, all subjects could successfully follow the instruction to maintain constant target muscle activation levels both for TA and TS muscles, which verifies the usefulness of the visual feedback provided.

The characteristic "peanut" shape stiffness structure that we found from our previous work [2, 9] was still evident in all muscle active conditions. One interesting result is that stiffness increased with muscle activation, but the increase was more pronounced in the DP direction than in the IE direction (Table I). This means that the ankle is still weak in the frontal plane even with voluntary muscle contraction. This result is consistent with the clinical observation that most ankle injuries, such as sprains and twisted ankles, occur in the frontal plane [17].

Although stiffness structures in active muscles were not just scaled-up copies of relaxed behavior, for any single movement direction in the coupled 2 DOFs, we found a highly linear relationship between muscle activations and stiffness values in that direction (Table II), at least under the conditions of this experiment.

Finding the relationship between muscle activation and ankle stiffness structure is important: if a convincing relation is constructed, we can predict ankle stiffness in any direction of movement of the coupled DOFs based on muscle activation level. Thus we may use a wearable robot not as a sensor to directly identify stiffness but as an actuator for other special purposes, such as subject training and rehabilitation.

In the current experimental setup, the maximum torque of the Anklebot (continuous stall torque) was sometimes not sufficient for the TS active study at 25% and 30% of MVC. To overcome this problem, we changed the input perturbation to utilize the instantaneous peak motor torque when needed.

In this study, we explored the relationship between muscle activation and ankle stiffness structure based on the identification of steady-state (time-invariant) dynamic ankle mechanical impedance. As a future direction, we plan to study the relationship between muscle activation and transient (time-varying) ankle mechanical impedance.

Acknowledgment

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