Smart Wearable Garment and Rapid Musculoskeletal Modelling for Accurate Neuromechanical Analysis

D. Simonetti, B. F. J. M. Koopman, and M. Sartori

Abstract Current clinical diagnostic tools for post-stroke motor deficits are based on rapid but subjective evaluation. Greater accuracy could be provided in well-equipped biomechanical laboratories. However, this involves lengthy set-up, data acquisition, and offline data analysis available only days, or weeks, after the initial assessment, no longer reflecting the patient's actual state. The ability to capture the patient's muscle activity as well as joint kinematics is key for neuromechanical assessment, but the process of muscle localization for electrode placement and joint angle measurement is not always viable in clinical environments. Here, we propose a new wearable technology with an integrated high-density electromyography and kinematics sensors able to provide simultaneously a rapid and accurate diagnosis. The soft sensorized garment in conjunction with automatic clustering of the HD-EMG channels into muscle groups and real-time modeling of the neuromusculoskeletal system allows estimation of internal mechanical forces in a rapid, quantitative, and clinically viable way.

1 Introduction

Currently, the clinical assessment of post-stroke gait is based on rapid but non-objective metrics; the functional ambulation categories (FACs) [1], i.e. self-paced 10-m walking speed and fastest 10-m walking speed. To provide a successful motor assessment, clinicians need to rapidly and objectively evaluate patient-specific neurophysical conditions over time. Electromyography (EMG)-driven musculoskeletal modeling and simulation [2] could potentially provide quantitative assessment but requires lengthy procedures, i.e. capturing the patient's myoelectrical activity as well as joint kinematics. This is already possible in biomechanical laboratories. Nevertheless, the high cost, the lengthy set up such as the process of manual muscle localization for electrode placement, joint angle measurements, and the time-consuming

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processing of the data are not viable in the clinical environment on a large group of patients.

The present work proposes an advanced technology that combines (1) a fully wearable soft sensorized garment, (2) an automatic algorithm for muscle localization from HD-EMG, and (3) a framework for patient-specific neuro-mechanical modeling (as shown in Fig. 1). The sensorized garment, in combination with automated muscle localization, allows reducing the set-up time and preventing human error to manually identify the muscles and place the electrodes. To automatize the process of clustering the HD channel space into muscles' specific activation, we apply non-negative matrix factorization (NNMF) [3] to extract muscle synergies during locomotion. We applied the NNMF-based extraction of muscle synergies to segment the HD-EMG grid capturing the spatial distribution of lower leg muscles without any prior knowledge on the electrodes' position. The NNMF is applied to HD-EMG to extract 4 main clusters related to the Tibialis Anterior (TA), Soleus (SOL), Gastrocnemius Lateralis (GL), and Medialis (GM), and Peroneus (PR). The average activation profile of each electrodes' cluster drive, thereafter, the framework for neuromuscular simulation of the patient-specific internal neuro-mechanical system.

In the following paragraphs, we show two preliminary studies assessing (1) extraction of 5 mean activation profiles from NNMF-muscle synergies using the soft HD-EMG garment placed on the lower leg and (2) torque estimation at ankle joint during different locomotion tasks using EMG-driven musculoskeletal model.

2 Material and Methods

Both experiments were performed on healthy male subjects performing locomotor tasks at three different speeds (slow, comfortable, and fast). Ground reaction forces (GRF), trajectories of 35 retro-reflective markers on the subject's body, and electromyography were recorded. In Test 1 the muscle activity was recorded on the right lower leg using a garment equipped with 64 monopolar electrodes equally distributed. Whereas, in Test 2, bipolar electrodes were used to record 9 muscles on the leg.

Test 1 focuses on the automatic extraction of muscle activation profiles from HD-EMG grid. The channel-space of 64 electrodes is reduced in 4 signals, one for each main muscle in the lower leg. This reduction is the output of the NNMF algorithm. To automatically identify the 5 lower leg muscles, the NNMF is performed two times. The first application of NNMF extracts conventional synergies describing the recruitment of two main muscle groups, calf, and TA, as a function of the gait cycle. The TA activation is computed as an average of the active channels that exceed a 70% threshold. However, the calf can still be dimensionally reduced in the remaining ankle muscles, Gastrocnemius Medialis and Lateralis, Peroneus, and Soleus. Therefore, the NNMF is applied a second time on the calf primitive active channels to identify 4 different groups of electrodes that activate together because part of the same muscle group. The synergies combined with the prior knowledge of muscle location in the leg allows associating each extracted sub-region to each muscle and extract and

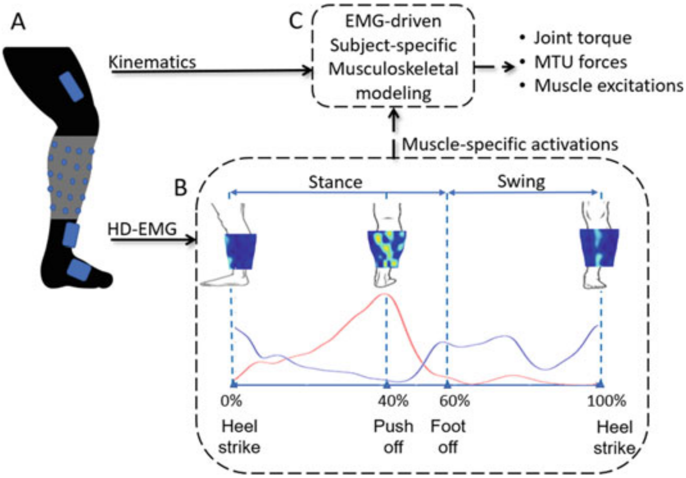

Fig. 1 Schematics of the proposed technology. **a** Fully wearable soft sensorized garment, **b** the automatic algorithm for the segmentation of HD-EMG. The two curves represent the conventional synergies during the push-off (red curve) and the swing (blue curve) phase. **c** Framework for patient-specific neuro-mechanical modeling

average activation. The computed mean activation profiles are then compared with the linear envelope of electrodes manually selected from the HD electrode grid that were located on each muscle belly.

Test 2 output is a subject-specific representation of the underlying neuromechanical processes obtained performing two steps: (1) modeling and simulation of the recorded movements and (2) simulation of musculoskeletal properties of the lower leg. Initially, the markers trajectories are input to the OpenSim software [4] to scale the musculoskeletal geometry and match the subject-specific anthropometry. Secondly the scaled model with the GRFs and the markers trajectories are processed in OpenSim to retrieve joint angles, joint torques, and musculotendon units (MTUs) lengths, and moment arms. The step (2) allows estimating subject-specific neuromechanical properties during dynamic motor tasks. First, the subject-specific model is calibrated to match the subject-specific internal parameters describing MTUs activation and contraction. Then the EMG-driven estimation of ankle torque is computed and compared with the experimental torque using the square of the Pearson coefficient of correlation (R^2).

3 Results

The 64 EMG channels are reduced by NNMF in 2 non-negative factors with 64 weightings each. The heatmaps of the weights showed two main active clusters: a big area that is supposed to be the calf and related to the push-off primitive, and a reduced cluster highlighting the TA active area related to the swing primitive. The second NNMF applied to the calf results in 4 synergies that cluster the grid in 4 different active regions. The extracted averaged activation profiles from these clusters resemble the ones from the manual selected channels with greater accuracy for 1 km/h ($R^2 = 0.97 \pm 0.03$) and 3 km/h ($R^2 = 0.95 \pm 0.06$) speeds respect to the faster locomotion velocity ($R^2 = 0.78 \pm 0.21$).

Test 2 instead assessed the ability to estimate joint torque from experimental EMG and inverse kinematics (IK) angles. Ankle torque estimation curves fit the relative inverse dynamics (ID)-moment with greater accuracy, for slow ($R^2 = 0.87 \pm 0.13$) and comfortable ($R^2 = 0.91 \pm 0.07$) walking speeds than the faster velocity ($R^2 = 0.63 \pm 2.6$).

4 Discussions

This work proposed a new technology that integrates advanced signal processing and musculoskeletal modelling techniques within wearable technology. We showed positive preliminary results of a new approach for lower leg muscle localization using NNMF-based muscle synergies extraction during locomotion and ankle joint torque estimation using bipolar EMG-driven musculoskeletal modeling offline. Merging these two steps in the future would lead to a new pipeline for fast and advanced clinical assessment.

With Test 1 we proved that (1) the NNMF can be used to localize active regions of the lower leg during locomotion and (2) we can associate them to specific muscles. The extracted activation profiles show to resemble the activation of all the lower leg muscles suggesting that the NNMF-based extraction of muscle synergies can be a suitable approach for the HD-EMG muscle localization.

The calibrated EMG-driven model can predict with good accuracy muscle-generated torque at the ankle joint in motor tasks not used for the subject model calibration. Using bipolar electrodes on muscles we can provide insights on subject-specific internal mechanical properties. Can we use HD-EMG to drive the estimation of internal mechanical properties with the same accuracy?

Future works will merge both experiments resulting in a framework for HD-EMG extraction of muscle-specific activation driving subject-specific neuromuscular modeling for rapid neuromechanical analysis.

5 Conclusion

The combination of flexible sensorized garment, the automatic procedure of muscle activity extraction added to the framework for neuromuscular modeling has a good potential to become a resource for clinicians. It will give them rapidity and a new and enhanced perspective on the patient's musculoskeletal system helping to tailor the rehabilitation treatment on the patient-specific needs for optimal recovery.

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