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Analysis of muscle synergies and activation–deactivation patterns in subjects with anterior cruciate ligament deficiency during walking

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ABSTRACT

Background: The knowledge of muscle activation patterns when doing a certain task in subjects with anterior cruciate ligament deficiency could help to improve their rehabilitation treatment. The goal of this study is to identify differences in such patterns between anterior cruciate ligament–deficient and healthy subjects during walking.
Methods: Electromyographic data for eight muscles were measured in a sample of eighteen subjects with anterior cruciate ligament deficiency, in both injured (ipsilateral group) and non-injured (contralateral group) legs, and a sample of ten healthy subjects (control group). The analysis was carried out at two levels: activation–deactivation patterns and muscle synergies. Muscle synergy components were calculated using a non-negative matrix factorization algorithm.
Findings: The results showed that there was a higher co-contraction in injured than in healthy subjects. Although all muscles were activated similarly since all subjects developed the same task (walking), some differences could be observed among the analyzed groups.
Interpretation: The observed differences in the synergy components of injured subjects suggested that those individuals alter muscle activation patterns to stabilize the knee joint. This analysis could provide valuable information for the physiotherapist to identify alterations in muscle activation patterns during the follow-up of the subject's rehabilitation.

1. Introduction

The rupture of the anterior cruciate ligament (ACL) is one of the most common knee injuries. It affects around two million people worldwide every year (Renström, 2013). Subjects with ACL deficiency alter their muscle activations when doing a certain task due to the lack of ACL. It is believed that muscles are activated synergistically following a certain pattern depending on the motor task (Lacquaniti et al., 2012; Ting, 2007; Ting and McKay, 2007; Ting et al., 2012), that is to say, our central nervous system (CNS) does not activate the muscles independently. Muscle synergies are represented by modules consisting of one neural command (NC), which represents the time activation of a set of muscles, and one synergy vector (SV), which represents the weighting factor of each muscle to its NC (Ting and Macpherson, 2005). The number of NCs is lower than the number of muscles. Therefore, the analysis of this lower dimensional activation pattern may explain the changes in neuromuscular activity due to the ACL rupture.

It is believed that the number of synergies used by a human being when walking is between 4 and 6 (Allen and Neptune, 2012; Clark et al., 2010; De Groote et al., 2014; Ivanenko et al., 2004, 2005; Oliveira et al., 2014). The variance accounted for (VAF) between the reconstructed and the original signals is evaluated to select the proper number of modules to be used when factorizing the signals. Most authors consider that a VAF > 0.9 is the threshold to accept the reconstruction (Clark et al., 2010; Oliveira et al., 2014). It is reported that there are similarities in the muscle synergies when performing the same movement across subjects. Several authors reported muscle synergies when walking (Clark et al., 2010; Dominici et al., 2011; Ivanenko et al., 2004; Neptune et al., 2009; Oliveira et al., 2014), walking with perturbations (Ivanenko et al., 2005) or performing other tasks (De Rugy et al., 2013). Clark et al. (2010) applied the muscle synergy analysis in post-stroke injured subjects. They observed that, although the patterns were similar among groups, the complexity in post-stroke injured subjects was lower than in healthy subjects, i.e., they needed fewer modules to have a good signal reconstruction. It is unclear what synergistic strategy is followed by joint-injured subjects to activate the muscles spanning that joint. Depending on the joint injury, subjects can apply different activation strategies to avoid pain or to stabilize the joint.

Apart from the clinical evaluation of muscle co-contraction, the use of the factorization can be useful for motion analysis and simulation. There is indeterminacy when calculating the muscle forces, since they

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cannot be calculated experimentally due to invasiveness. The usual method to estimate the forces is with the resolution of an optimization problem (Erdemir et al., 2007), which consists of minimizing a cost function (a physiological variable) that represents the strategy of the CNS to activate the muscles. The optimization results can produce multiple physiologically feasible solutions due to the muscle redundancy. Some authors used the muscle synergy components to decrease the indeterminacy in the muscle force calculations, either in forward dynamics (Allen and Neptune, 2012; Neptune et al., 2009) or inverse dynamics (Walter et al., 2014) approaches. Regarding subjects with ACL deficiency, differences have been observed at the joint level as well as at individual EMG signals (Houck et al., 2007; Knoll et al., 2004; Rudolph et al., 2001; Serrancolí et al., 2014). As far as the authors know, the muscle synergy analysis has not yet been applied to subjects with this kind of injury. In consideration of that, this study could be useful at two levels. On the one hand, in a clinical application, it would allow the specialist to follow the rehabilitation process of injured subjects. On the other hand, in a motion dynamic analysis, muscle synergies could be used to decrease the indeterminacy in the muscle force calculation of subjects with ACL deficiency.

The main goal of this study is to evaluate and compare the muscle activation patterns in healthy and injured subjects during walking. In particular, the analysis is carried out at two levels: activation–deactivation patterns and muscle synergies. In our study, all subjects with an ACL deficiency were considered adapters (Button et al., 2006) and the measures were done a few days or weeks before the surgery of the ligament reconstruction. Although muscle synergy patterns can present many similarities among groups, since all of them perform the same task, human gait, our hypothesis was that the pattern of muscle synergy components may have different tendencies. As mentioned, there are studies that evaluate individual muscle activations in subjects with ACL deficiency, but the objective of this study is to evaluate the differences in muscle synergies compared to healthy subjects in order to better understand the muscle activation pattern in absence of ACL function. The knowledge of the differences in muscle synergies for subjects with ACL deficiency could help a physiotherapist to redirect the rehabilitation treatment. The analysis comprises two steps. The first is a comparison of the activation–deactivation pattern among healthy legs (control group), injured subjects' injured legs (ipsilateral group) and injured subjects' non-injured legs (contralateral group). Then, a muscle synergy analysis is reported and compared among the three groups.

2. Methods

2.1. Subjects

Ten healthy subjects, five men and five women (mean (SD): age 31.5 (12.9) years, mass 65.2 (7.6) kg, height 170.4 (8.6) cm), and eighteen subjects with ACL deficiency, twelve men and six women (mean (SD): age 32.3 (10.99) years, mass 68.5 (9.7) kg, height 172.1 (7.6) cm), volunteered as participants in this study. No healthy subjects suffered any lower-limb injury. The injured subjects were classified as adapters, according to the medical staff and the widely used classification presented in Button et al. (2006), which considers that they can be divided into three groups: copers, who return to the preinjury level of their daily tasks and sport activities; non-copers, who cannot return to their preinjury level of tasks and sport activities and have episodes of full giving way even in daily tasks; and adapters, who reduce or modify certain tasks or the sport level to prevent their knee giving way. All injured subjects reported that they could deal with daily life and they did not suffer pain when normal walking; however, they felt discomfort and pain when they did sports that required knee pivoting, such as football or skiing. The time interval from the injury varied from one month to three years (mean (SD): 10.3 (12.0) months). All subjects provided their consent to contribute to this study.

2.2. Experimental setup

All volunteers were asked to walk a minimum of three overground gait cycles at a self-selected speed (mean (SD): 0.77 (0.12) m/s for healthy subjects and 0.80 (0.13) m/s for injured subjects). One of the gait cycles was selected from the recorded trials and was analyzed.

EMG data from sixteen muscles were measured with sixteen surface EMG sensors (Biometrics, Newport, United Kingdom) at 1000 Hz. The signal of eight lower-limb muscles from each leg of the subject was measured (tibialis anterior, TA; soleus, SO; gastrocnemius lateralis, GL; gluteus maximus, GM; rectus femoris, RF; vastus lateralis, VL; semitendinosus, ST; and extensor digitorum longus, ED). These muscles are the main contributors to human walking. The EMG data for each subject (right and left leg) came from the same gait trial, which decreases the variability due to the differences that could appear when measuring different gait trials separately.

EMG signals were demeaned, rectified and filtered with a Butterworth low-pass filter at 6 Hz. Then, they were normalized by maximum voluntary contraction (MVC) values obtained by MVC exercises that were previously done. The exercises were selected to calculate the maximum muscle excitations (Kendall et al., 2005). The volunteers were asked to apply force against a resistance along a direction to activate the muscles responsible for ankle plantar flexion/dorsiflexion (ED, SO, TA, GL), knee flexion/extension and abduction/adduction (RF, ST and VL) and hip flexion/extension and abduction/adduction (GM, ST and RF). Data from these trials were processed in the same way that walking trials (demeaned, rectified and filtered at 6 Hz). The maximum values of EMG were selected among all available trials (MVC exercises and gait trials). These values were verified visually and individually in each subject to avoid the acceptance of a wrong maximum value. All MVC exercises, as well as verifications, were carried out by the same technician to standardize the comparison. Using this normalization, the signal was constrained to be between 0 (not activated) and 1 (maximum activation). So, an activation close to 1 would mean that the muscle is near its maximum activation.

Ground reaction forces (GRF) and marker trajectories were also measured to identify the events of the gait cycle. The GRF were measured by means of two force plates (AMTI, Watertown, MA) at 100 Hz. Two marker trajectories from each foot (heel and tip of the first metatarsal bone) were captured by fourteen infrared cameras (Naturalpoint, Corvallis, OR). Once the gait cycle was identified for each leg, data was interpolated to 101 frames. Normalized EMG data are available on the net as supplementary data.

2.3. Data analysis

Data analysis was carried out by means of MATLAB R2010a (Mathworks, Natick, MA). All data were divided into three groups: control, which consists of data from healthy subjects; ipsilateral, from the ipsilateral leg, which is affected by the ACL injury; and contralateral, from the non-injured leg of the subjects with ACL deficiency.

2.3.1. Activation–deactivation patterns

An initial analysis of the activation–deactivation pattern for each muscle was carried out to identify the differences in the activation timing between groups. The onset–offset activation pattern was calculated for each subject, considering EMG signal to be activated when it was higher than the following threshold:

$$Threshold_{on-off} = \min(EMG) + 0.5(\max(EMG) - \min(EMG)) \quad (1)$$

where EMG stands for an EMG signal. The activation pattern was calculated for each group. A muscle was considered to be active when more than 50% of the subjects had this muscle activated at a particular time frame.