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Changes in synergy of transtibial amputee during gait: A pilot study

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Abstract— The number of lower limb amputations is increasing significantly in developed countries. The knowledge of muscle synergy in subjects with loss of muscles could help to understand the general neural strategy underlying muscle coordination in walking. The aim of this study was to investigate the differences in healthy subject's dominant leg, amputee's intact leg (IL) and the amputee's prosthetic leg (PL) muscles using synergy analysis. Concatenated non-negative matrix factorization (CNMF) was performed to divide the surface electromyography (sEMG) data obtained from 6 upper knee and 4 shank muscles into muscle synergy (S) and activation coefficient profile (C) during walking. The difference in S showed low to high correlations inter-subjectively. The high correlation suggests that the central nervous system (CNS) activates the same groups of muscles synergistically. Amputee's muscle alterations due to inadequate proprioceptive feedback, weight bearing deficiency in PL and prosthesis type could lead to a low correlation in S between groups. The C showed to be statistically significantly different in some regions of the gait cycle (GC). These findings could provide valuable information for rehabilitation purposes and development of a synergy-based controller from sEMG for future generations of prostheses.

I. Introduction

The prevalence of lower limb amputation has been rising rapidly with the primary causes associated with dysvascular disease and traumatic injuries. This number is expected to increase to one in 95 Americans in the next 35 years [1]. In developed countries, the most common type of amputation is transtibial amputation. The gait characteristics of the unilateral transtibial amputee (TA) subjects change considerably to that of healthy subjects (HS) in terms of gait mechanics and muscle coordination patterns [2]. Fay et al. [2] reported, the prosthetic leg (PL) muscle activity in the hamstring and vasti muscles were prolonged as compared to the intact leg (IL) and non-amputee. Increase in activation and delays in peak activations have been observed in prosthetic knee flexor muscles from early to mid-stance. In the IL, the increased activation has been noticed in early stance and late swing phases. The activation patterns in healthy leg and amputee's IL have been shown to be similar, whereas prosthetic limb muscles showed greater variation in knee muscles [2]. The PL biceps femoris long head, vasti and rectus have been shown to increase activity from early to mid-stance than IL [2]. It is believed that the central nervous system (CNS) controls the muscles in a few low-dimensional muscle synergies (modules). Factorization techniques such as principal component analysis and non-negative matrix factorization (NNMF) have been implemented to identify synergy components. Each of these components consists of synergy vector (S), which represents relative weighting of each muscle within the corresponding synergy, and activation coefficient profiles (C), which represents the timing of synergy vector being active over time. Previous studies reported that the number of synergy groups is between 4 and 6 when walking [3-5]. The number of synergy groups can be identified by means of variance accounted for (VAF), which evaluates the correlation between sum products of the synergy components (i.e. reconstruction) and original signals. The VAF > 0.9 has been accepted as a standard threshold for reconstruction [3, 4]. NNMF have been implemented in a large variety of terrains and locomotion conditions such as walking in different conditions including forward and backward, different speeds, perturbation, under different loading condition, curvilinear, uphill and downhill slope [4-6]. In the studies associated with walking, a considerable difference has been shown in the synergy vector. However, an insignificant change was observed in activation profiles. Several studies implemented NNMF for comparison of healthy and pathological populations [4, 5, 7, 8]. Clark et al. [7] reported, lower number of synergy groups were required to achieve a high VAF in post-stroke patients compared to the healthy population. Serrancoli et al. [8] found, anterior cruciate ligament (ACL) deficiency would cause a slight change in activation profiles of ACL patients compared to HS. In our previous works [4, 5], we investigated the difference in muscle synergy components between healthy and elderly TA subjects during ramp ascending and descending. The results showed a reasonable correlation between S, however, revealed significant difference in C at some regions of the gait cycle (GC) inter-subjectively. To the best of our knowledge, the comparison of synergy analysis has not yet been made on TA during walking. In this study, we aim to evaluate and compare the S and C between young HS and an elderly TA during walking. Four different cases have been investigated: Case 1) HS' dominant leg versus amputee's IL (10 muscles); Case 2) HS' upper knee (UKN) versus amputee's UKN IL (6 muscles); Case 3) HS' UKN versus amputee's UKN PL (6 muscles); and Case 4) amputee's UKN IL versus amputee's UKN PL (6 muscles). As it was reported, there have been studies, which investigated the individual muscle activation in terms of patterns, duration, and co-activity in TA subjects, but the objective of our study was to elucidate the difference in a low dimensional scale. We hypothesized that the muscle synergy components are significantly different in cases 2-4. The information in this study could help in rehabilitation of amputee subjects as well as the development of a future generation of myoelectric prostheses. In addition, the

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differences in activation profiles of this group will help the neurophysiologists to better understand the shift in synergy and complexity of TA muscle recruitment strategy by CNS.

II. Procedure

A. Experimental Protocol

Four active male HS (mean (SD): age 21.3 (0.4) years, weight 72.2 (5.9) kg, height 175.7 (6.0) cm) and one active elderly TA (age: 76; weight: 69.3 kg; height: 185.1 cm) volunteered as participants. The HS' dominant leg, amputee's IL and PL were free of any lower limb injury, any skin condition or neurodegenerative disease except the amputation of the amputee's PL. Before the experiment, the subjects were informed of the experimental procedure, and consent was obtained. The Institutional Review Board approved the experimental procedures involving human subjects described in this study. The subjects were asked to walk at their self-selected speed. sEMG was recorded from the HS' dominant leg, amputee's IL and PL muscles using Noraxon sEMG electrodes (Telemyo, Noraxon, USA) at 1500 Hz. An ultrasound scanner (LogiQ e, GE Healthcare, USA) was utilized to attach the sEMG electrodes to the belly of ten muscles i.e. HS' dominant leg and amputee's IL (case 1). In addition, six UKN muscles of amputee's PL was considered (cases 3 and 4). Among the recorded muscles, six were from UKN including rectus femoris (RF), biceps femoris long head (BFLH), tensor fascia latae (TFL), gluteus medius (GMED), and vasti (i.e. vastus medialis (VM) and vastus lateralis (VL)), and four was from below knee including; anterior tibialis (AT) and triceps surae (i.e. soleus (SOL), gastrocnemius medialis (GASTMED) and gastrocnemius lateralis (GASTLAT)).

III. Algorithm Description

Concatenated non-negative matrix factorization (CNMF) approach was implemented to linearly decompose the concatenated data into a linear combination of S and C [4, 5, 9]. The norm-2 length of S was constrained to be 1 to minimize the indeterminacy of the factorization. A matrix A^c (n×m) was created to combine and concatenate the sEMG data, where n represents the # of subjects × # of GCs × 101 and m accounts for the # of muscles. The results of CNMF populated: (1) the concatenated coefficient (C^c = n×k) and (2) the fixed muscle synergy (S = k×m), where k represents the number of synergy groups. The summation of the product of muscle weightings in all the synergies and their corresponding coefficients provides the reconstructed signals.

A. CNMF Frameworks

Initially, random values of C and S were chosen. In order to obtain optimal C and S, an alternating least squares algorithm was implemented in the framework. These values must satisfy the Frobenius norm to minimize the error J=||A - CS||$_F$. Perturbation was introduced to the solution and used as a new initial guess to ensure optimality of the results. To find the final optimal solution of S and C, three iterations were performed for the whole framework. However, the perturbation was not applied in the last iteration [4, 5, 9].

B. Data Analyses

Initially, sEMG signals were demeaned and rectified. To obtain the linear envelope of the signals, a low pass filter (zero lag 2nd order Butterworth at 6 Hz) was performed. The data was normalized to the maximum value calculated from each muscle over all selected trials. Therefore, all values obtained from each muscle were ranged between zero and one. Data were interpolated to 101 points to represent one GC. In order to find the appropriate number of muscle synergy groups in healthy and amputee groups and adequately reconstruct the original signals, concatenated variance accounted for (VAF) was calculated. The formula below shows the calculation of VAF;

$$\mathrm{VAF} = (1 - \sum_{o=1}^{n}\sum_{p=1}^{m} e_{op}^2 \,/\, \sum_{o=1}^{n}\sum_{p=1}^{m} A_{op}^2) \qquad (1)$$

Where e is the error, i.e., A - CS, and the indices o and p represent the rows and the columns of the quantities e and A. The VAF > 0.9 has been accepted as a standard threshold for determining the appropriate number provided that an extra synergy improves the VAF for less than 0.05. We performed a functional sorting because each group might use muscle synergies differently and subsequently to facilitate inter-subjective comparison. This method resolves the large differences in contribution to the total data variability by rearranging the indices of synergy and coefficient of one group based on the other group. Consequently, in cases 1-3, HS' synergy components were chosen as references to sort the amputee's results and in case 4 the amputee's IL as a reference to sort amputee's PL. Intra-class correlation (ICC) was done to investigate the reconstruction quality of each muscle intra-subjectively by assessing the shape and pattern of the reconstructed signal. ICC and $R^2 < 0.5$, $0.5 <$ ICC and $R^2 < 0.75$, and ICC and $R^2 > 0.75$ imply low, moderate, and high correlation, respectively. The comparison between the S and C between the groups was made by means of the coefficient of determination (R^2) and statistical parametric mapping (SPM), respectively.

IV. Results and Discussions

The similarity of reconstructed and original signals was calculated by the formula (1). The total VAF showed to be higher than 0.9 when four (case 1) and three (cases 2-4) synergy groups were chosen. In cases 2-4, three synergy groups satisfied the VAF criteria. Fig. 1 shows changes in a number of synergy groups in all four cases by means of VAF.

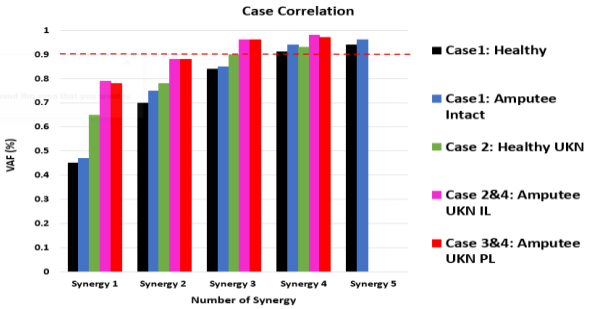

Figure 1. VAF comparison as a function of number of synergies in cases 1-4. In Fig. 1; HS' dominant leg (Case 1; black), amputee's IL (Case 1; blue), HS' UKN (Case 2; green), amputee's UKN IL (Case 2 and 4; magenta), amputee's UKN PL sorted with respect to HS's UKN and amputee's IL (Case 3 and 4; red) were illustrated as color codes, respectively.

An individual muscle signal reconstruction showed a reasonable correlation to the original signal in all four cases (see Fig. 2). The lowest and highest ICC illustrated in RF

(Case 2: healthy) and AT (Case 1: amputee) with a correlation value of 0.31 and 0.99, respectively.

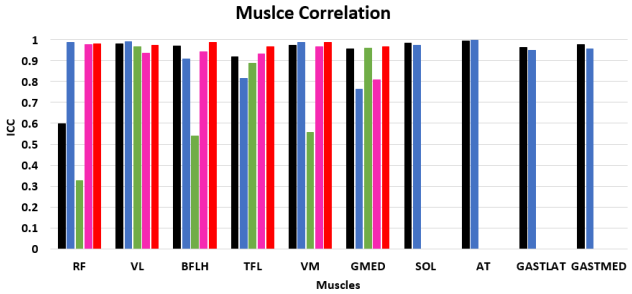

Figure 2: ICC of reconstructed signals compared to the original signal for each muscle in cases 1-4. Refer to Fig. 1 for colour code explanation.

Fig.3 A and B shows S and C of cases 1-4, respectively. The muscle synergy is divided into two muscle groups based on their weighting; primary > 0.5 and secondary < 0.5 (Fig.3 A). The subsequent section is the interpretation of each case. Refer to Table 2 caption for gait cycle phases explanation.

Case 1: In HS, synergy 1 (S1) showed a high activation of vasti, TFL and GMED muscles as well as a lesser activity of RF, SOL, and AT during the ES and TW. Synergy 2 (S2) consisted of primarily triceps surae muscles and to a lesser extent TFL and GMED during the MS-TS. Synergy 3 (S3) composed of primarily AT muscle and GMED and RF as secondary muscles during the ES and IW-TW. Synergy 4 (S4) comprised of BFLH as primary and vasti, RF and AT as secondary muscles during the ES and TW.

Case 1: In amputee's IL, S1 was composed of mainly vasti high activation and lower activation of TFL, BFLH GMED, SOL, and AT during the ES-MS and TW. S2 consisted of triceps surae and to lesser extent TFL and BFLH during the MS-TS. S3 composed of primarily AT muscle during the ES-MS and whole swing phase. S4 consisted of RF as a primarily with the highest activation along with a lesser activity of TFL, VM, and AT as secondary muscles during the ES-MS and IW.

Case 2 and 3: In HS' UKN, S1 showed a high activation of GMED, and lesser activity of RF, VM, and BFLH during the MS-TS. S2 consisted of mainly vasti muscles and to a lesser extent RF and TFL muscles during ES-MS and IW. S3 composed of high activation of TFL, and lower activation of BFLH, VL, RF and GMED during ES-MS and TW.Case 2 and 4: In amputee's UKN IL, S1 was comprised of primarily RF high activation and lower activation of VM during the stance phase and IW. S2 consisted of mainly vasti and BFLH muscles and lesser activity of GMED and RF during ES-MS stance and TW. S3 showed a high activation of TFL as primary muscle and low activation of VL, GMED, and BFLH as secondary muscles during the whole stance phase.

Case 3: In amputee's UKN PL, S1 showed a high activation of TFL and GMED muscles and to a lesser extent VM and RF muscles during the ES. S2 consisted of mainly vasti and RF muscles as primary and lower activation of GMED during ES-MS and TW. S3 was composed of primarily BFLH muscle and lesser activity of GMED, VM and RF as secondary muscles during the whole stance phase and TW.

Case 4: The interpretation of amputee's UKN PL is similar to the case 3 but the order of synergy group 1 and 3 switches.

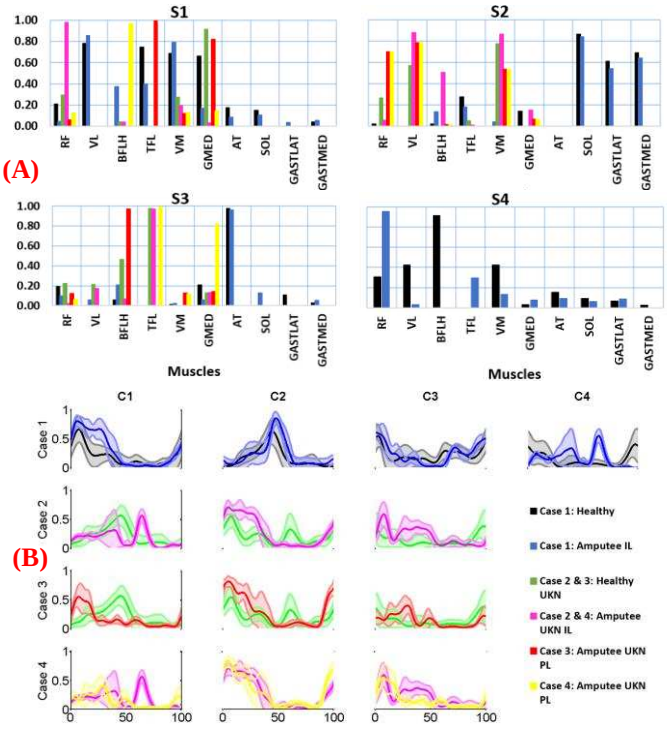

Figure 3. Muscles weighting within each synergy (A) and activation profiles (B) of cases 1-4. Refer to Fig. 1 caption for colour code explanation. Note; amputee UKN PL sorted with respect to amputee IL shown (Case 4; yellow). In Fig. 3 (B), the thick line and shaded area show the mean and one standard deviation values of C for cases 1-4, respectively.

To assess the difference in the S and C of each case, R^2 and SPM were done, respectively (Table 1 and Fig. 4).

Table 1: Muscle synergy comparison in case 1-4 using R^2

	S1	**S2**	**S3**	**S4**
Case 1	0.74	0.97	0.92	0
Case 2	0	0.57	0.84	NA
Case 3	0	0.70	0	NA
Case 4	0	0.57	0.48	NA

Table 1 shows S correlation in all cases. In case 1, the highest and lowest correlation occurred in S2 and S4, respectively. Case 2 showed a low, moderate and strong correlation in S1, S2 and S3 between the two groups, respectively. In case 3, low correlation of S1 and S3 and moderate correlation of S2 were observed. In case 4, S1 showed low correlation, but S2 and S3 revealed moderate correlation.

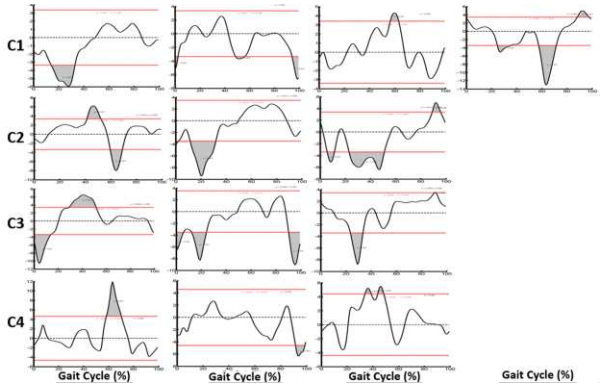

Figure 4. SPM analysis using a two-sample t-test. The regions in the GC where $p < 0.05$ (grey shaded area) are shown in all the four cases.

In the SPM analysis, the t-value is zero when there is no difference between the mean C of the two groups. When it is positive; meaning $C_{healthy} > C_{amputee}$ and when it is negative; vice versa. The critical value is calculated through inference based on random field theory. If the t-curve exceeds the threshold (red line), then this shows the statistically significant difference in C. Table 2 shows the regions where t curve passed the t-critical values of all four cases in Fig. 4.

Table 2. Difference in mean activation profiles of case 1-4. Early, mid and terminal stance represent ES, MS, and TS; initial, mid and terminal swing represents IW, MW, and TW, respectively.

P < 0.05				
	C1	**C2**	**C3**	**C4**
Case 1	MS*	ES, TS, TW(*)	TS	MS*, IW*-TW
Case 2	LS, IW*	ES, MS(*)	ES*, MS-TS*, TW	NA
Case 3	ES*,MS-TS	ES, MS, TW(*)	MS*, TW	NA
Case 4	IW	TW*	MS, TS	NA

* and (*) indicate coefficients associated with amputee's IL (Case 1 and 2) and amputee's PL (Case 3 and 4) is significantly greater in one region and more than one region compared to the other group within their respective case, respectively.

In this study, the differences between the synergy components of healthy and amputee subjects were investigated. The VAF results revealed a similar number of synergy between the two groups in all the cases. This indicates the complexity acquired by CNS to control the muscles of both groups in each case is similar. The increase in the number of synergy groups resulted in a higher VAF. However, the optimal number was chosen based on the VAF criteria. In addition, we checked if the optimal group number provided better distinct synergy groups and physiological relevance of muscle EMG contents. In order to compare the S of all cases, R^2 was performed. In case 1, triceps surae had the highest weight in S3 indicating a strong correlation between HS and TA. However, S4 showed a low correlation due to the difference in the primary muscle of the HS (BFLH) and TA (RF). The weighting of the primary muscle showed to have a significant effect on the correlation of the healthy and amputee in terms of S. The lowest correlation was found in case 3 where S1 and S3 showed no common primary muscles between HS' UKN and amputee's UKN PL. The highest correlation was found in case 1 where the first 3 synergy groups (S1-S3) had the strongest correlation between HS' dominant leg and amputee's IL. The R^2 in case 4 showed a moderate correlation of S2 and S3 and a low correlation of the S1 between amputee's UKN IL and UKN PL. The results in case 1 revealed a reasonable correlation between muscle synergy (S1-S3) suggesting CNS synergistically activates the same group of muscles in both groups. The low to moderate correlation in some synergy groups (Case 1: S4; Case 2: S1 and S2; Case 3: S1 and S3); Case 4: S1-S3) could be due to weight bearing deficiency and inadequate proprioceptive feedback of the PL as well as type of prosthesis (mechanically passive) which could result in muscle synergy alteration. The results of the t-test in SPM showed activation of the two groups, regardless of high correlation in S, could be different significantly in some regions of the GC. Case 1 showed amputee's C1 is significantly different from HS at 14% - 32% of the GC. The largest significant differences have been observed in case 1 (C4), case 2 (C3), case 3 (C1) and case 4 (C1). This indicates C controlling the muscle synergy is not similar in terms of timing and level of activations in multiple regions of the GC in all cases. This alteration in the behavior of the muscles could be due to the loss of limb, which causes higher GRF in the IL, as well as prosthesis properties (e.g. stiffness), which result in greater mechanical work on both IL and PL. In addition, sarcopenia (loss of muscle strength and volume) could play a major role in modifying the muscle activation of elderly TA. The potential limitations of this study were the low number of participants and the age difference between them.

V. Conclusions

In this study, muscle synergy analysis was used to elicit functional changes in TA. It was revealed that TA's CNS complexity to recruit muscles is analogous to those of non-amputees. In each case, the primary muscle(s) had a significant impact on the level of S correlation. The C results suggested that amputee's IL and PL were significantly different when compared together and to the HS. These findings can provide useful information to therapists to tailor rehabilitation methods in TA. Also, this study will help the prosthetic companies to develop synergy-based prostheses.

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