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# Muscle Synergy Analysis in Transtibial Amputee during Ramp Ascending Activity *

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**Abstract—In developed countries, the highest number of amputees are elderly with transtibial amputation. Walking on inclined surfaces is difficult for amputees due to loss of muscle volume and strength thereby transtibial amputees (TA) rely on the intact limb to maintain stability. The aim of this study was to use the concatenated non-negative matrix factorization (CNMF) technique to calculate muscle synergy components and compare the difference in muscle synergies and their associated activation profiles in the healthy and amputee groups during ramp ascending (RA) activity. Healthy subjects' dominant leg and amputee's intact leg (IL) were considered for recording surface electromyography (sEMG). The muscle synergies comparison showed a reasonable correlation between the healthy and amputee groups. This suggests the central nervous system (CNS) activates the same group of muscles synergistically. However, the activation coefficient profile (C) results indicated statistically significant difference ($p < 0.05$) in some parts of the gait cycle (GC) in healthy and amputee groups. The difference exhibited in activation profiles of amputee's IL could be due to the instability of the prosthetic leg during the GC which resulted in alteration of the IL muscles activations. This information will be useful in rehabilitation and in the future development of prosthetic devices by using the IL muscles information to control the prostheses.**

## I. Introduction

Previous research showed that the highly complex tasks are performed by a few low dimensional muscle synergies [1, 2]. The synchronization of the muscles activations forms muscle synergies or modules [2]. In order to detect activation of the muscles, surface electromyography (sEMG) can be used. The high dimensional sEMG signals can be decomposed into two components: (i) muscle synergy which represents relative weighting of each muscle within the corresponding synergy and (ii) activation coefficient profile which represents the timing of muscle synergy being active over time i.e. gait cycle (GC) [2]. The human body uses different control strategies to appropriately recruit lower limb muscles when the ambulation is changed.

One of the main activities of daily livings is walking on the inclined surfaces i.e. slope ascending/descending. The importance of these tasks become more prominent when pathological groups such as amputees are considered in the study. According to Vicker et al. [3], in the developed countries, the large number of people who lost their limbs are elderly transtibial amputees (TA). Study of amputee's intact leg (IL) muscles is as crucial as prosthetic leg muscles due to their significant roles in maintaining stability during different activities [4]. Vickers et al. [3] found, in ramp ascending (RA) and descending, the lateral hamstrings muscle of IL were active throughout the GC. In addition, vastus lateralis was also active for the most part of the GC except during mid-swing (SW) phase. It was shown in the same study that in the IL the plantar flexion of ankle does not occur at the end of the stance (ST) phase in order to compensate for the weight transfer onto the prosthetic leg. This will result the toes of the IL to balance the entire body weight in both activities. The IL of the amputees showed increase in amplitude in the knee muscles than healthy subjects [3].

According to Lay et al. [5], in healthy young and old adult populations, there was a considerable increase in recruitment of hip, knee and ankle extensor muscles during uphill walking as compared to the level walking. However, in old population, either due to muscle strength or neural modification, the reliance on the hip muscles are greater than the ankle muscles [5]. Despite the difference in the EMG activation pattern for each muscle, burst of activity is one of the common features between different muscles [1]. Factorization analyses including non-negative matrix factorization (NNMF) have been implemented to identify muscle activation patterns in synergies and activation profiles [1]. Such analyses resulted in the identification of 4-6 synergies during activities such as walking and running [6]. The appropriate number of synergies can be determined by calculating the variance accounted for (VAF) between the reconstructed and the original signals. The acceptable VAF has been reported to be above 0.9 [6]. NNMF has been applied in order to find muscle synergies in different movements including forward and backward walking, walking with different speeds [1], walking with perturbation, walking under different loading condition, curvilinear walking, running, cycling and pedaling as reported in [7]. In all the studies associated with walking, a very insignificant change has been shown in the activation profiles however, a considerable difference has been illustrated in the muscle weightings. Other groups applied the NNMF technique to investigate the difference between healthy and pathological populations [6, 8]. Clark et al. [6] reported, in post-stroke subjects, similar activation profiles with fewer synergies were required to have a high VAF during walking when compared to healthy subjects. Serrancoli et al. [8] found that patients

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with anterior crucial ligament deficiency have slightly different muscle synergy profiles during walking comparing with healthy subjects. In the above literatures, there have been studies that investigated the individual muscle activation in healthy young and old adult as well as elderly TA during uphill activity.

However, in this study, we aimed to investigate the difference between young healthy and old TA muscles using muscle synergy analysis methodology in order to have a better understanding of the muscle synergy and activation profile when the limb is missing. This information will be useful in rehabilitation and also in the future development of prosthetic devices in order to control the prosthetic leg when the information is obtained from the IL by means of various sensors such as sEMG. To the best of authors' knowledge, muscle synergy analysis has not yet been applied to EMG of TA subjects during RA. We hypothesized that muscle synergies and activation profiles of the amputee's IL are significantly different to those of healthy subjects.

## II. Procedure

### A. Experimental Protocol

Four active healthy male subjects (age: 21.3 ± 0.4 years old; 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) were participated in this study. The healthy subjects and amputee's IL were free of any lower leg injuries, any skin condition or neurodegenerative disease. The experimental procedure was explained and consent form was obtained from the subjects prior to the experiment. The experimental procedures involving human subjects described in this study were approved by the Institutional Review Board. Noraxon sEMG electrodes (Telemyo, Noraxon, Scottsdale, USA) were used to collect electrical signals unilaterally from the dominant leg of the healthy subjects and IL of the amputee. Subjects performed RA on an instrumented inclined walkway (5°) at a self-selected pace. The sEMG was placed over the rectus femoris (RF), vastus lateralis (VL), biceps femoris long head (BFLH), tensor fascia latae (TFL), vastus medialis (VM), anterior tibialis (AT), soleus (SOL), gluteus medius (GMED), gastrocnemius lateralis (GASTLAT) and gastrocnemius medialis (GASTMED). In order to detect the muscle bellies, electrode placements were done by a specialist using an ultrasound scanner (LogiQ e, GE Healthcare, USA). sEMG was sampled at 1500 Hz and amplified by a gain of 1000.

## III. Algorithm description

In this study, concatenated non-negative matrix factorization (CNMF) approach was used to linearly decompose the concatenated data into a linear combination of muscle synergies (S) and concatenated coefficients ($C^c$) [9]. In order to minimize the indeterminacy of the factorization, the S was constrained to be 1. The sEMG data were combined and concatenated into matrix $A^c$ (n-by-m), where n represents the # of subjects × # of GCs × 101 and m represents the # of muscles. Using CNMF, two matrices were populated: (1) the concatenated coefficient ($C^c$ = n-by-k) and (2) the fixed muscle synergy (S = k-by-m), where k represents the number of synergies (= 4). The reconstruction of EMG signal is calculated by summation of the product of muscle weightings in all the synergies and their corresponding coefficients.

### A. CNMF Frameworks

Random values of C and S were chosen in order to initiate the CNMF. An alternating least squares algorithm was implemented to attain optimal C and S. These values must satisfy the Frobenius norm to minimize the error $J=||A - CS||_F$. To ensure optimality, perturbation was introduced to the solution and used as new initial guess. 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 (adopted from [9]).

### B. Data Analyses

A band pass filter (zero-lag Butterworth filter with a cut off frequency of 20-500 Hz) was used to remove motion artefacts and high frequency noise from the signals. In order to obtain the muscle activation pattern (linear envelope), full wave rectification and low pass filter (zero lag 2nd order Butterworth at 6 Hz) were performed. The data was normalized to the maximum peak value obtained from each muscle over all selected GCs. Therefore, all values of each muscle were ranged between zero and one. The data was then interpolated to 101 data points corresponding to the instrumented leg GC. In order to compare healthy and amputee data sets, functional sorting method was implemented. This method rearranged the indices of synergies and coefficient of one group based on the other group. The sorting in this study was done by choosing the healthy subjects' synergy and coefficient vectors as reference to sort the amputee results. Concatenated VAF ($VAF^c$) was then calculated for the concatenated data ($A^c$) to find the appropriate number of synergies required for a task. The $VAF^c$ was defined for each group as follows;

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

Where e is the error, i.e., A—CS, and the indices o and p represents the rows and the columns of the quantities e and A. Intra-class correlation (ICC) was performed to investigate the reconstruction quality of the results intra-subjectively. The shape and pattern of the reconstructed signal was compared to the original signal. According to the literature [9], ICC < 0.5, 0.5 < ICC < 0.75, and ICC > 0.75 imply low, moderate, and high correlation, respectively. A two-sample statistical parametric mapping (SPM) t-test was chosen to calculate the p-value.

## IV. Results and discussion

All the muscles have been reconstructed using the muscle weightings and activation profiles with a total $VAF^c$ greater than 90% in both healthy and amputee groups during RA. $VAF^c$ was defined to analyze the similarity of the reconstructed EMG with the original signal [9]. A comparison study was done on the number of required synergies in previous literatures such as [6]. In this study, 4 and 5 muscle synergies were chosen to find the optimal number of synergies in healthy and amputee groups during RA. Healthy subjects showed slightly higher $VAF^C$ (= 0.95) using 5 synergies as compared to the $VAF^c$ (= 0.93) using 4 synergies. Similarly in the amputee, 5 synergies showed a little increase in $VAF^c$ than 4 synergies (0.97 and 0.95, respectively). Both healthy and

amputee groups had a high correlation, by means of statistical analyses, (0.9 < ICC and $R^2$), between the reconstructed and the original signals of all the muscles except RF in which moderate correlation has been perceived. In healthy subjects, synergy 1 (S1) showed a high activation of BFLH, vasti muscles and moderate activation of RF and GMED in the early ST phase and end of SW phase. Synergy 2 (S2) composed of plantar-flexor muscles high activation and to a lesser extent GMED, TFL, BFLH and RF activations during the mid to end-ST of the GC. Synergy 3 (S3) consisted of activity in early to mid-ST primarily from GMED and TFL and lesser activation can be perceived from vasti muscles, AT and SOL. Synergy 4 (S4) consisted of AT muscle prominently active in early ST, early SW and terminal SW (Fig. 1 a & b). In the amputee subject, S1 was composed of BFLH high activation and less activity of vasti and plantar flexor muscles except GASTLAT during early ST, mid-ST and terminal SW. S2 consisted of activity in mid-ST till early SW phase primarily for plantar flexors and TFL muscles and lesser activity in RF and GMED muscles. S3 showed a high activation in vasti muscles and GMED and to a lesser extent in TFL, BFLH, SOL and GASTMED muscles in early ST to mid-ST. S4 consisted of AT with the highest activation along with a lesser activity of RF, GASTLAT during early ST and the whole SW phase (Fig. 1 c & d).

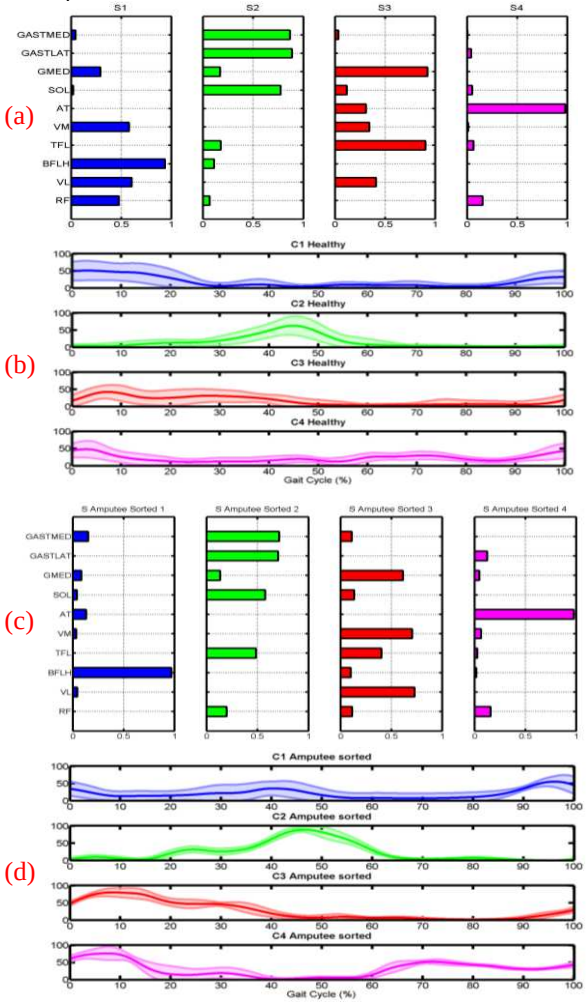


Figure 1. Muscles weighting of each synergy (a & c) and their associated activation profiles (b & d) of the healthy and amputee groups. In Fig. 1 (b and d), the thick line shows the mean value of the activation profile with the shaded area representing one standard deviation.

After functional sorting had been performed, the inter-subject analysis was done using the coefficient of determination ($R^2$) and SPM. $R^2$ was implemented to compare the results of muscle synergies between the two groups (see Table. 1).

Table 1. Comparison between the muscle synergies of healthy and amputee groups by means of $R^2$

| **Synergy** | **S1** | **S2** | **S3** | **S4** |
|---|---|---|---|---|
| **$R^2$** | 0.52 | 0.90 | 0.66 | 0.98 |

In order to investigate the difference between the activation coefficient profiles of healthy and amputee groups, a two-sample SPM t-test was chosen to calculate the p-value. The p-value < 0.05 was considered as statistically different. Fig. 2 (left) shows the mean activation coefficient profiles and the shaded standard deviation clouds in healthy and amputee groups. In Fig. 2 (right), t-value continuum (or t-value trajectory) is illustrated. The horizontal black dashed line indicates the mean difference between the two groups is zero. The area above and below the mean (t = 0) is known as H and A, respectively. In addition, the grey areas on the right where the curve passed the t-critical value (red dashed line) indicated statistically significant difference between the two groups.

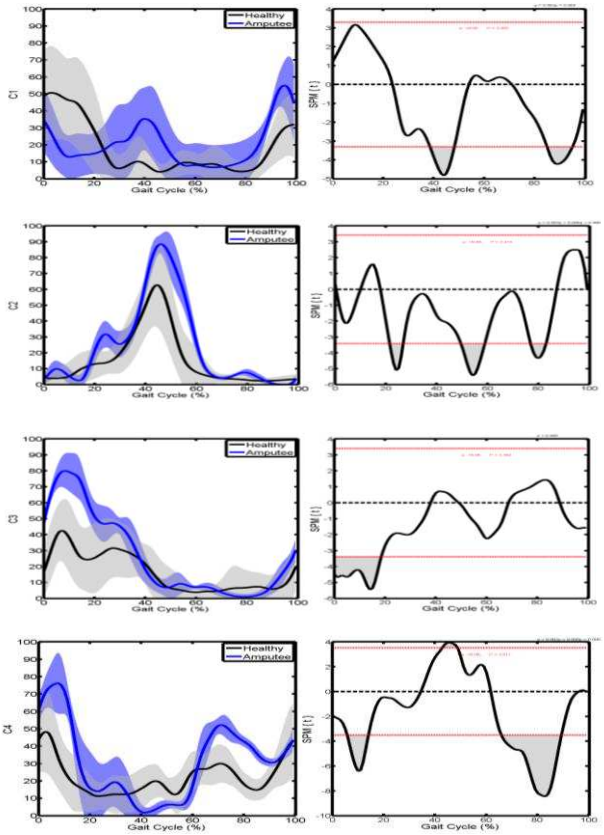


Figure 2. SPM analysis using t-test; the figures on left shows the C for healthy (grey) and amputee (blue). Refer to Fig. 1 for meaning of thick, shaded area. The plots on the right illustrate the statistically significant difference area (shaded) between the two groups. The horizontal black dotted dashed line indicates t=0. The red dashed lines show the t critical threshold value.

In Fig. 2 (right), A > H (below t = 0) and A < H (above t = 0) indicate a greater and lower activation profiles mean difference in amputee as compared to healthy subjects during

the GC, respectively. In addition, t-values above the upper threshold (t-critical) suggest significantly greater healthy values, and t values below the lower threshold suggest significantly greater amputee values. Table 2 shows the comparison between the mean activation profiles of the two groups.

Table 2. The difference in mean activation profiles of healthy (H) and amputee (A). The numbers written in the bracket represents the percentage of the GC (%) where the mean is higher (A > H) and lower (A < H) for the amputee. t-critical illustrates statistically significant different between the groups.

| | **A > H (%)** | **A < H (%)** | **P < 0.05 (%)** |
|---|---|---|---|
| **C1** | (25-38), (49-54) (72-84), (95-100) | (0-24), (55-71) | (39-48), (85-95) |
| **C2** | (0-10), (18-21), (27-49), (59-76), (84-86) | (11-17), (87-100) | (22-26), (50-58), (77-83) |
| **C3** | (20-37), (49-68), (90-100) | (38-48), (69-89) | (0-19) |
| **C4** | (0-5), (15-29), (62-65), (90-96) | (30-42), (50-61), (97-100) | (6-14), (43-49)*, (66-89) |

* indicates healthy activation profile is significantly greater than amputee

In this study, the difference between the healthy and amputee groups was determined by investigating the muscle synergies and activation profiles. The number of synergies were similar in healthy and amputee groups which suggested that the complexity of muscle synergy are analogous due to similar control strategy used in the central nervous system (CNS). This study supports the findings of the [8] which analysed subjects with ACL deficiency. The $VAF^c$ analysis showed a higher $VAF^c$ using 5 synergies however, 4 synergies were believed to provide more distinct synergy groups of muscle EMG contents. Moreover, since both $VAF^c$ values were higher than 0.9 and the difference between them was not more than 0.05, 4 synergies were chosen for both groups.

The difference between the muscle synergies and activation profiles of healthy and amputee groups was done by means of statistical analysis ($R^2$ and SPM). The $R^2$ analysis on muscle synergies showed values of 0.52, 0.90, 0.66 and 0.98 for the S1, S2, S3 and S4 inter-subjectively, respectively. The results revealed a reasonable correlation between the muscle synergies of the two groups. A high correlation in S2 and S4 suggests the CNS activates the same group of muscles synergistically in both groups. A moderate correlation in S1 and S3 showed slight change in muscle weightings. This indicated amputee's IL was subject to small muscle synergy alteration may be due to lack of force tolerance, inadequate proprioception of the prosthetic leg or stiffness of the prosthetic device.

In order to investigate the difference in the activation profiles of the two groups, the two-sample SPM t-test was performed (Fig. 2). The results indicated that the coefficient of two groups are significantly different at some percentage of the GC. Comparing C1 in healthy and amputee groups showed two significant differences in the activation profiles (39%-48% and 85%-95%) during the GC. C2 showed differences at 22%-26%, 50%-58% and 77%-83% of the GC. C3 only showed one area to be significantly different which occurred during the early to mid-ST phase (0%-19%). Healthy subjects' C4 was statistically significantly greater than amputee subject at 43%-49% time whereas amputee's C4 was significantly different than healthy subjects at 6%-14% and 66%-89% of the GC (see Fig. 2 and Table. 2).

In this study, there are several limitations that need to be considered. The age difference could result in high variation in the outcome. Although, the TA was an active individual, sarcopenia which results in loss of muscle mass and strength could impact muscle activation during RA. Another limitation is the type of prosthesis that was used by the amputee. This may hinder the walking ability which results in larger amount of mechanical work on the IL especially in inclined surfaces. Therefore, adaptation to such devices could alter the behavior of the muscle synergy activation in the IL.

## V. Conclusions

The study presented CNMF as a robust algorithm which facilitates the comparison of muscle synergy analysis between populations by keeping synergies fixed between subjects. Despite the fact that the IL of the amputee was free of any pathologies, the results showed there was a significant difference in activation profiles of the amputee's muscles compared to the healthy subjects. The limitations of this study include age difference, sarcopenia and type of prosthesis in amputee. This work will be beneficial for future development of the myoelectric prostheses and rehabilitation. Further work will include implementation of muscle synergy analysis with higher number of subjects, different types of amputees and for various activities.

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