An Investigation of HDEMG Spatial Parameters in Individuals with Transtibial Amputation

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Abstract— **Mutlichannel surface electromyography (EMG) or high density surface EMG (HDsEMG) can be used to examine spatial activity of muscle in those with amputation. HDsEMG has not been used extensively to study muscle adapations of those with transtibial amputation and this tool can be particularly valuable to study residual muscle activity after amputation. Torque and HDsEMG were recorded from the rectus femoris muscle during isokinetic knee extensions of varying speed in able-bodied participants (n=2) and those with transtibial amputation (n=2). Spatial distribution was estimated using the RMS value for each of the 64 electrode grid locations and 2-Dimensional (2D) maps were developed for each participant. Peak torque, and HDsEMG spatial features were compared across speed and the clinical participants were compared with an age-matched control participant. The results from this exploratory study showed that HDsEMG spatial parameters can be used to provide insight regarding muscle function, particularly those representative of muscle heterogeneity. This study also showed that HDsEMG can be successfully used with those with lower limb amputation. The information obtained can be used to help improve the design of powered limbs and training protocols.**

Keywords— **High density electromyography, prosthetics, spatial activity, transtibial amputation**

I. INTRODUCTION

Lower limb amputation can occur due to complications from diabetes mellitus or other vascular diseases, trauma, congenital deformities, or bone cancer [1]. Currently, more than 44 000 Canadians have undergone a lower limb amputation. Of these, 13 000 have gone through a transtibial amputation, which is currently considered the most common form of amputation in Canada [1].Transtibial amputation can have a negative impact on the individual's body image, vocation, and ability to socialize [2] and may interfere with an individual's ability to perform activities of daily living. In addition, associated changes in the anatomical structure can lead to asymmetries in the musculature of the lower limbs leading to alterations in the individual's gait, and reductions in their activity level.

The risk of developing secondary complications after a transtibial amputation suggests there should be more focus on performance assessments of the lower limb musculature during rehabilitation. Furthermore, muscle strength is an important determinant of functional performance and lower limb muscles are critical for locomotion and activities of daily living. Examining dynamic contractions with varying joint angles can provide insight into lower limb mechanics. Isokinetic dynamometry allows the measurement of dynamic contractions in a controlled speed and enables researchers to examine dynamic movement in a controlled environment.

Isokinetic dynamometers can be used to safely measure muscle strength for both dynamic and static conditions [3] and can be used effectively with those with lower limb amputations.

Currently powered prostheses use the electrical activity in the residual muscle as a controller, however few studies have characterized the muscle activity generated by that residual muscle in the lower limb. Surface electromyography (EMG) uses electrodes placed over the muscle bell to record the electrical activity that produces skeletal muscle contraction. Parameters of the resulting EMG signal can provide insight regarding muscle physiology. The challenge with traditional surface EMG is that it captures gross muscle activity rather than individual motor units (MU) and while intramuscular EMG can be used to examine MU behaviour, it can be challenging for clinical populations. MU behaviour is necessary for greater understanding of muscle performance during dynamic activities and therefore to facilitate accurate prostheses control. More recently, multichannel two-dimensional grids of electrodes, or "high density" EMG, have been used to evaluate muscle function successfully [4-8]. These high density surface EMG (HDsEMG) recordings can be used to evaluate the detailed properties of MU activity [4]. In addition, the spatial distribution of HDsEMG can be used to examine alterations in MU behaviour [6]. Previous studies have used HDsEMG to show that spatial activation distribution in a muscle is non-uniform and that the spatial distribution pattern can be altered by contraction levels or fatigue [4-7]. The advantage of HDsEMG is that it can be used to non-invasively investigate MU activation of muscle during force production at varying levels of force contractions [8] making it desirable for clinical populations. Few studies have used HDsEMG to examine amputated musculature and to our knowledge there are no studies that have used HDsEMG to investigate dynamic lower limb activity in those with transtibial amputation.

The purpose of this study was to compare strength and recruitment patterns of the rectus femoris during knee extension and flexion between individuals with and without transtibial amputation. It was hypothesized that individuals with transtibial amputation will demonstrate decreased strength and differences in recruitment patterns in the rectus femoris compared to individuals without transtibial amputation. In addition, it was expected that there would be differences between the affected and unaffected leg in those with transtibial amputation. The results from this study will help to better characterize muscle activity in both the intact and residual limb of those with lower limb amputation.

II. Methods

A. Participants

A control group of 10 healthy participants and two clinical participants with transtibial amputation were recruited to participate in this study. The control group comprised four males and six females, with a mean age of 38.4 ± 18.4 years old. All individuals identified their right sides as dominant. The two clinical participants included one 23-year-old male (Clinical 1) with a right transtibial amputation and one 77-year-old female (Clinical 2) with a left transtibial amputation. Both participants required transtibial amputation due to trauma and had been using their prostheses for approximately 7.5 years prior to testing. Both participants identified the side opposite to the side with the amputation as their dominant side. Both participants were able to ambulate independently without the use of crutches or a cane.

Residual limb length, measured using a tape measure from the tibial tuberosity to the tip of the residual limb, had an average residual limb length of 8.7 ± 0.4 cm. Residual limb circumference, measured using a tape measure at the distal end of the residual limb, had an average residual limb circumference of 26.8 ± 1.8 cm.

B. Experimental Design

Prior to testing, participants were assessed using the following criteria: (1) ambulatory without a mobility aid; (2) absence of injury in the last six months; (3) absence of health complications that would create a risk of injury during the study; and (4) absence of inflammation and pain of the knee joint. All participants were given an overview of the data collection equipment, procedures, and information on any risks associated with testing. Participants were asked to read and sign an informed consent form provided by the experimenter. This research study was approved by the University of New Brunswick's Research Ethics Board.

The participants were asked to complete a questionnaire regarding their age, date of birth (DOB), date of amputation, reason for amputation, activity level, chronic conditions, medical history, and the "Get Active" questionnaire. Preliminary measurements were taken, and the prostheses were examined.

C. Experimental Protocol

Each participant was asked to walk around the track for five minutes at a comfortable speed to warm up their muscles. Upon returning to the laboratory, the participant was asked to sit in the chair of the isokinetic dynamometer. The rectus femoris muscle was palpated on both sides to determine the appropriate location for the HDsEMG 64-grid. The skin area was shaved and swabbed with alcohol, and the HDsEMG 64-grid was placed. Since the clinical population did not have an ankle on the amputated side, the patella was palpated, and the ground electrode was placed on the patella for all participants. The isokinetic dynamometer was then set up for the participant (Figure 1). All changes such as the back angle, as well as dynamometer height, were noted if they were different from the standard set-up. The fulcrum of the dynamometer was aligned to the rotational axis of the knee using the lateral femoral condyle as the bony landmark [9]. The length of the fulcrum was then recorded. Participants were secured in the chair of the isokinetic dynamometer with straps placed firmly around their chest, and the knee extension adapter was secured approximately two inches above the ankle joint. The range of motion was measured as well as the weight of the limb. The isokinetic dynamometer was set at a velocity of 0°/sec for the isometric contraction with a knee angle of approximately 90°. Once the set-up was complete, the participant was asked to perform one maximum voluntary contraction (MVC) that was held for a duration of five-seconds, followed by a break (three minutes for the control group and four to five minutes for the clinical group). Participants were then provided an opportunity to practice isokinetic submaximal contractions. Participants were then asked to produce isokinetic contractions randomly at three contraction velocities slow (60°/sec), moderate (90°/sec), and fast (120°/sec) speeds. The participant was encouraged to "push as hard as you can" throughout the contraction. For each of the three speeds the participant completed two maximal contractions with rest between the contractions. While isokinetic dynamometry is well established for able-bodied individuals, a reliability test was completed for both clinical participants by repeating the assessment on both sides two weeks later and showed similar results.

Participants were tested at the Human Performance Lab located in the Richard J. Currie Center at the University of New Brunswick using the Cybex HUMAC®/NORM™ Testing and Rehabilitation System. Data was recorded at a frequency of 100 Hz. Additionally, muscle related variables were collected using a Multichannel HDsEMG device (Sessantaquattro, OT Bioelecttronica, Italy). HDsEMG data was recorded at a frequency of 1024 Hz, using a semi-disposable 64-channel electrode grid positioned over the rectus femoris muscle. The grid was composed of five columns and 13 rows of electrodes. Then, a double-sided adhesive foam grid was set on the electrode grid, a conductive cream was spread and into each cavity to create skin contact with each electrode. A pre-gelled ground electrode (Duo-Trode, MyotronicsIncorporated, Washington, USA, diameter: 12.5mm, material: Ag/AgCl) was positioned on the patella.

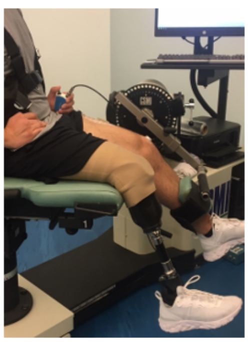

Fig. 1 Clinical participant during testing

D. Data Analysis

HDsEMG data was recorded using the product software (OT Biolab, Bioelettronica, Italy). EMG data was filtered with a bandpass filter 20-400Hz and the Root Mean Square (RMS) was calculated for every one-second interval for each channel (OT Biolab). EMG data were analyzed using a two-second interval centred at the midpoint of the contraction.
Spatial distribution was estimated using the Root Mean Square (RMS) EMG signals for each of the electrode grid locations from which 2D maps were developed (Figure 2). The 2D maps represent the intensity of muscle activation over the surface of the muscle [10]:

$$HM_{ij} = RMS\ (sEMG_{ij}) \qquad (1)$$

where HM is an activation map and each pixel in a map (HM_{ij}) corresponds to an RMS value of a channel in an electrode array (position i,j).[10] Entropy and coefficient of variation (CoV) were used to represent the increased variety of signals (heterogeneity) of the HDsEMG distribution. When all of the channels show the same RMS values, maximum entropy occurs. When all of the channels except for one have an RMS value of zero, minimum entropy occurs. Therefore, as entropy increases, heterogeneity of the muscle fibre types decreases.

$$E = -\sum_{i=1}^{59} p(i)^2 \ log_2\ p(i)^2 \qquad (2)$$

Where the square of the RMS value at electrode i, *p(i),* is then normalized by the sum of the squares of all 59 RMS values [4].

CoV was calculated by taking the standard deviation (SD) of the 59 RMS values divided by the average of the 59 RMS values. The CoV value is small when the SD is small relative to the mean which occurs when the channels are more uniform. A smaller CoV value implies low heterogeneity and therefore homogeneity of the muscle fibers. All the signal processing and evaluation was performed using a custom designed MATLAB software [11].

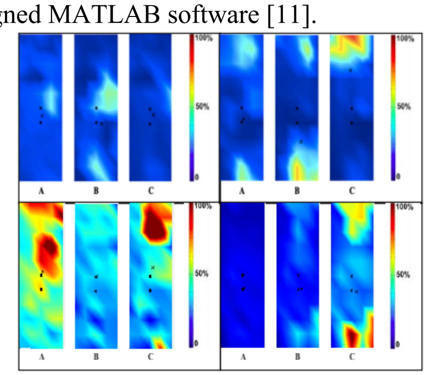

Fig 2: Clinical 2 and Age Matched Control During Isokinetic Contractions. Top left panel indicates heat intensity maps of Clinical 2 of the sound (right) rectus femoris during the following three velocities. A) 60°/sec; B) 90°/sec; C) 120°/sec. Top right panel indicates heat intensity maps of Clinical 2 of the affected (left) rectus femoris. Lower left panel indicates age matched control data for the dominant (right) rectus femoris and lower right panel indicates age matched control data for the non-dominant (left) rectus femoris.

III. RESULTS

This study was limited to two clinical participants and therefore only descriptive statistics are provided. Table 1 presents the data for each clinical participant comparing their affected side (with amputation) to the non-dominant leg of their age-matched control. Both the clincal and control participants demonstrated decreasing strength with increasing speed (Table 1). The mean RMS decreased with increasing speed for both clinical participants, however it increased with speed for the control participants. Clinical 1 (young male) showed similar entropy across speeds compared to the control participant. Clinical 2 and their age-matched control had similar strength results however Clinical 2 showed lower entropy at the slower speeds suggesting greater muscle heterogeneity compared to the age-matched control participant. Clinical 2 also presented higher CoV than the the control participant suggesting greater heterogeneity. This participant was also the older female (age = 77 years old) suggesting that age may be a factor.

Table 1: Peak torque and spatial parameters of clinical participant (prosthesis side) and their age-matched control (non-dominant side).

	Clinical 1	Age-matched Control	Clinical 2	Age-matched Control
Peak Torque (Nm)				
60°/sec	100.3	154.3	40.8	40.6
90°/sec	85.4	102.0	26.0	28.8
120°/sec	73.2	82.5	30.9	24.8
Root Mean Square (mV)				
60°/sec	0.120	0.087	0.187	0.028
90°/sec	0.107	0.102	0.084	0.084
120°/sec	0.101	0.142	0.086	0.071
Entropy (au)				
60°/sec	5.32	5.39	2.14	5.52
90°/sec	5.40	5.44	3.21	4.25
120°/sec	5.38	4.97	4.56	4.63
CoV (au)				
60°/sec	48.0	46.5	155.4	38.5
90°/sec	43.0	44.6	115.2	101.0
120°/sec	43.6	63.4	76.3	83.4

IV. CONCLUSION

This study showed that HDsEMG can be used to non-invasively examine the spatial distribution and neuromuscular physiology of those with amputation. The results from this exploratory study showed that spatial parameters can be used to provide insight regarding muscle function, particularly those representative of muscle heterogeneity. The older female prosthesis user demonstrated differences in both entropy and CoV compared to the control participant suggesting greater muscle heterogeneity. While limited in sample size, this preliminary data suggests that further investigation is warranted using HDsEMG features for this population, in particular with respect to aging muscle.

More comparative studies with a larger number of participants and testing sessions are needed to create a better understanding of the effect of transtibial amputation on the musculature of the lower limb in order to improve rehabilitation methods and long-term outcomes.

Conflict of Interest

The authors declare that they have no conflict of interest.

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