**Development and assessment of a method to estimate maximum voluntary contraction EMG from submaximal EMG data**

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**Running Title:** Neck EMG MVICs estimated from SVCs

**Abstract**

EMG normalization (often to maximum voluntary isometric contraction (MVIC)) is used to control for inter-participant and day-to-day variations. Repeated MVIC exertions may be inadvisable from participants safety perspective. This study developed a technique to predict the MVIC EMG from submaximal isometric voluntary contraction (SVC) EMG. On Day 1, ten participants executed moment exertions of 100%, 60%, 40% and 20% of maximum (biceps brachii, rectus femoris, neck flexors, and neck extensors) as EMG data were collected. On Day 2, participants replicated the joint moment values from Day 1 (60%, 40% and 20%), and also performed MVIC exertions. Using the ratios between MVIC EMGs and SVC EMG data values established on Day 1, and the Day 2 SVC EMG data values, the Day 2 MVIC EMGs were predicted. The average absolute percentage error between the predicted and actual MVIC EMG values for Day 2 were calculated: biceps brachii 45%, rectus femoris 27%, right and left neck flexors 27% and 33%, right and left neck extensors both 29%. There will be a trade-off between the required accuracy of the MVIC EMG and the risk of injury due to exerting actual MVIC. Thus, using the developed predictive technique may depend on the study circumstances.

**Keywords:** electromyography, normalization, safety, cervical

**Word Count: 4011 (including acknowledgements)**

## Introduction

The characteristics of an electromyographic (EMG) signal captured using surface electromyography is influenced by a number of extrinsic and intrinsic factors. De Luca[1] discussed several factors that can influence the collected EMG signal, including electrode positioning, configuration and orientation, blood flow in the muscle, and the amount of tissue between the electrode and muscle of interest. Other researchers have further identified perspiration and temperature[2] as well as cross-talk of surrounding muscles[3] as factors that influence EMG signal. These factors can vary among participants in an experiment and can also vary from day-to-day in a single participant, making it difficult to accurately compare experimental EMG signals across days and individuals. To overcome the challenges associated with the day-to-day and person-to-person EMG signal variability, researchers often collect EMG data during reference contraction(s) and then normalize the experimental EMG data relative to the EMG values generated during these reference contractions.

Different types of EMG reference contractions have been employed to create the "denominator" reference value of the normalization process. Some of these reference values include the EMG from the maximum voluntary contraction (MVC) (either isometric or non-isometric contractions)[4-11]; the EMG from a submaximal voluntary contraction (SVC) (either isometric or non-isometric contractions)[1,12,13]; or a value of the EMG derived from the experimental task that is evaluated[5,6,13-17]. There is no consensus on the best method for EMG normalization that would be effective in all research studies[18-21]. An MVC during an isometric contraction "maximum voluntary isometric contraction" (MVIC) is the most common reference value to normalize the EMG data[22] and is widely used for normalization in EMG-assisted biomechanical models[23, 24]. While the MVIC technique is widely used to normalize EMG data, it has faced criticism from researchers[6,13,25] including the verity of the maximum contraction, increased injury risk, and potential fatigue effects.

It is often not easy to obtain a true and reliable MVIC for EMG normalization[23], as it can be impacted by participants' motivation and sincerity[26,27] as well as participant's level of pain/discomfort[28]. Furthermore, utilization of the MVIC method can be restricted or impossible for individuals with musculoskeletal disorders[29], and the MVIC method can also be uncomfortable or cause injuries in healthy participants in regions vulnerable to injury[23,30,31]. One example is collecting MVIC values of the neck muscles. Maximal force exerted by the neck musculature can cause injury, particularly when the neck is in a non-neutral position[32], or when simply executing MVIC exertions that are not typical for the neck region. To further add to this challenge, the neck MVIC technique can be affected by the pain or the fear of pain in participants with a history of neck pain[33]. These challenges could be also troublesome in other regions with a history of an injury and susceptible to reoccurrence of the injury. Furthermore, in some situations, the participant is able to exert MVIC, but the maximum exertion may not be advisable on the day of data collection (e.g., surgeons before performing a surgical procedure). The development of a method to generate estimates of day-specific MVIC EMG values with a minimal number of true MVIC exertions would achieve the goals of normalization of experimental EMG data while improving the level of safety of the participants, especially in a typical multi day experiment.

Marras and Davis[26] proposed a normalization method for the lumbar musculature that did not require participants perform an MVIC exertion to generate the MVIC EMG. Their method involved developing regression equations to predict maximum trunk contraction moments based solely on anthropometric measurements. Then, sub-maximal and maximal EMG data from a new set of participants were used to develop a linear regression model to determine the EMG-moment relationship for each of the lumbar muscles under study. This relationship was extrapolated to the previously predicted maximum contraction moment to get an EMG normalization reference point. This method assumed there was a relationship between anthropometry and trunk moments. The authors reported a large portion of unexplained variability in this model, which indicates that the anthropometry-trunk moment model has limitations.

The purpose of the current study was to develop and assess a regression modeling technique that seeks to predict MVIC EMG based on a previously (different day) established relationship between that muscles MVIC EMG and an SVC EMG value. Following the example of Marras and Davis[26], we propose a linear regression model between MVIC EMG and SVC EMG values. Our regression equations will be based on the participant's own sub-maximal and maximal exertions. It is hypothesized that the relationship between SVC EMG and MVIC EMG on a reference day can be used effectively to estimate MVIC EMG on another day wherein only SVC exertions are performed. If found to be suitably accurate, this technique could be employed in studies that require multi-day participation and would result in an increase in participant safety (as MVIC EMG is collected only on the first participation day) while still reaping the benefits of the MVIC EMG normalization technique.

## Methods

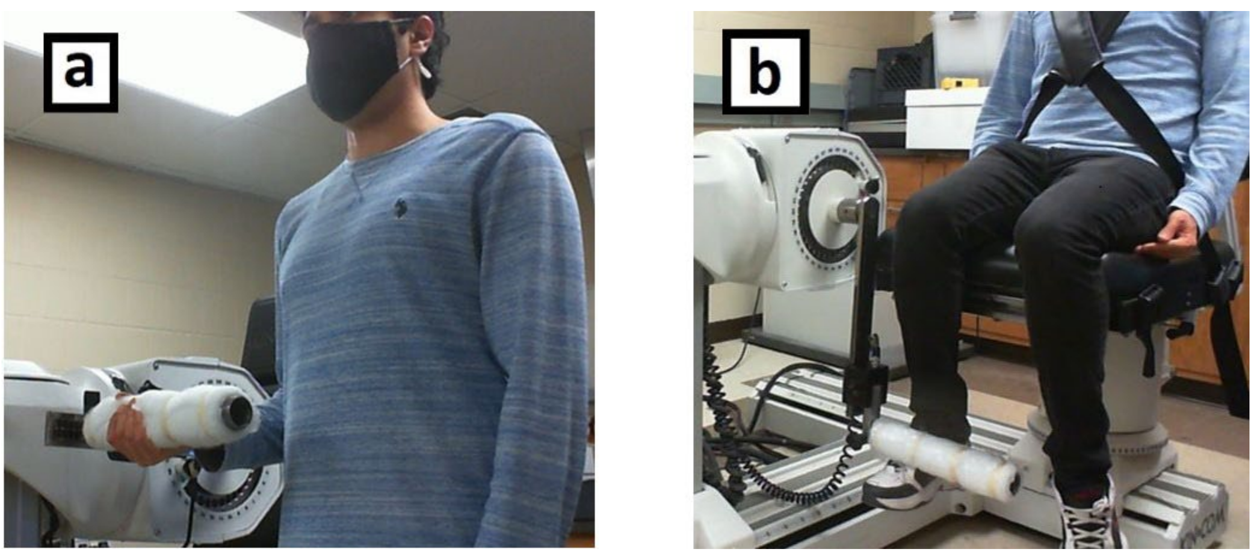
**Participants:** Ten healthy adults (nine males, one female) were recruited for this study. The mean and standard deviation (SD) of their age and select anthropometric measurements were as follows: age 34 (9) years, stature 175.3 (9.2) cm, body weight 77.5 (23.2) kg. All participants self-identified as right-handed (also considered as their dominant side).

**Data Collection Instrumentation:** The experimental apparatus consisted of two main instruments. A Kin-Com Isokinetic Dynamometer (125E, Chattanooga TN, USA) was used to provide isometric resistance while measuring moments generated by the muscles during maximum voluntary isometric contractions (MVICs) and submaximal voluntary contractions (SVCs). The posture of the joint under study during exertions was standardized and fixed using the dynamometer. Surface electromyography (EMG) was employed to collect the desired muscles activities using DELSYS Trigno Wireless Biofeedback System (Delsys Inc., MA). Each wireless sensor was 27x37x13 millimeters and weighed 14 grams. The surface EMG data were collected at a frequency of 2148 Hz. The dynamometer provided visual feedback so that the participant could observe the real-time generated moment (presented as a number on the video monitor) and match the joint moment requirements for a given SVC exertion trial.

**Experimental Procedure:** Each participant participated two times (different days) with at least a 48-hour interval between the two data collections to control for fatigue and carryover effects. The basic demographic and anthropometric data were collected on the first day of participation. On each day of data collection, the experiment was explained to the participant, and an informed consent was obtained. The participants were then guided through a series of nonstrenuous warm-up/stretch exercises that focused on the neck, elbow, and knee regions (flexion/extension, rolling and lateral motions of the neck, standing elbow flexion/extension, and standing hamstring curl). Then, the participant's skin at the location of the EMG electrodes was cleaned using alcohol. One surface EMG electrode was attached to the biceps brachii on the participant's dominant side, and one electrode was attached to their rectus femoris on the same side. SENIAM recommendations for sensor locations[34] were followed for biceps brachii and rectus femoris. Four EMG electrodes were attached to the participant's neck at right and left anterior locations and right and left posterior locations at the C4/C5 level of the cervical spine. The neck sensors were bilaterally symmetrical and inferiorly-superiorly oriented. The offset from the midline was 3.0-3.5 cm for the anterior sensors and 2.0-2.5 cm for the posterior sensors depending on the participant's neck width.

On each day of data collection, the MVIC and SVC exertions performed were 1) elbow flexion (biceps brachii electrode), 2) knee extension (rectus femoris electrode), 3) neck flexion (anterior neck electrodes), and 4) neck extension (posterior neck electrodes), respectively. The biceps brachii and rectus femoris were included to explore the generalizability of the technique developed in this study. The experimental procedure followed for each of these muscle groups was the same, so while the full procedure is described here for the biceps brachii, all described experimental procedures were likewise followed for the other three muscles/muscle groups. The only difference was the arrangement of the dynamometer and participant’s position at the dynamometer (Figures 1-2). The detailed description of the dynamometer set-up and participant’s position for each muscle/muscle group is available online (supplementary file 1). For the biceps brachii, the participant was asked to flex their elbow, exerting their maximum voluntary contraction force in the biceps brachii for two seconds. Participants were asked to relax their muscle and then get to their maximum exertion in a quasi-linear ramp increase in around two seconds overall and hold this maximum torque for two seconds as steadily as possible. Participants performed this exertion two times with one-minute rest between the two MVIC exertions. Verbal encouragement was used during all MVIC data collections. The momentary peak generated moment was observed and recorded by the experimenter using the visual feedback of the dynamometer, and the EMG electrodes collected the activity of the biceps brachii muscle. The greatest value of the peak generated moment during the two trials was selected as the maximum moment for the biceps brachii. The values of 20%, 40% and 60% of this moment value were calculated. After a two-minute rest interval from the completion of the second MVIC exertion, the participants were asked to use the video feedback system to generate 20%, 40%, and 60% of this

MVIC moment value with one-minute rest between the exertions. The order of presentation of the three levels of %SVC moment was randomized independently for each specific participant, muscle, and day of data collection. The participants were asked to exert the corresponding torques for the SVC exertions steadily with an acceptable error of about ±5%. They performed each SVC exertion only once unless the acceptable error was violated. In such cases, the participant was given one-minute rest and then the corresponding SVC exertion was repeated (only three occasions (different participants) throughout the study). As they performed these exertions the EMG data were collected and noted as 20%, 40%, and 60% of biceps brachii MVIC EMG. In each of these exertions the participant used the video feedback to hold the required moment for two seconds, and the muscle activities were collected at this steady state. The MVIC and SVCs of rectus femoris and neck flexor and extensor muscles were recorded following similar procedures. At the completion of the experimental trials, electrodes were removed, and the participant was led through a five-minute cool-down procedure and then they were free to go.



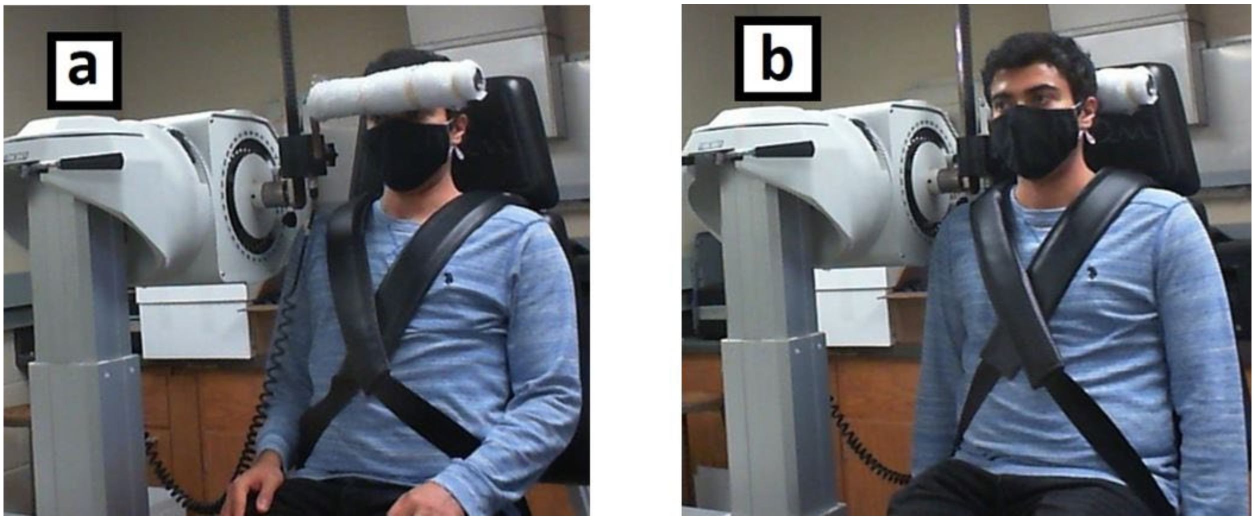



**Figure 2**: Apparatus and participant position (a)neck flexion & (b)neck extension exertions

On the second day of the data collection, the exact same procedure was repeated except that the SVC moments generated were those that were used on Day 1 (those that were calculated based on the MVIC moment of the first day of participation). For example, the SVCs of the biceps brachii on the second day were 20%, 40%, and 60% of the MVIC moment value of biceps brachii on the first day of data collection. New MVIC exertions were also collected per muscle for Day 2, independently; following a similar procedure used in Day 1. Identical EMG sensor placement procedures were followed on both days to ensure the consistency of the electrode location across the two days of the data collection. It was also ensured that on Day 2 the moment arm for exerting moments on the dynamometer were the same as Day 1 for each muscle and participant.

**Data Processing:** All EMG data were demeaned and filtered using a fourth order bandpass Butterworth filter (high pass=10Hz, low pass=400Hz) and a second order band-stop Butterworth filter (60 Hz and its closest harmonic alias (120 Hz)) (supplementary file 2). Then, the EMG data in time domain were rectified. The rectified MVIC EMG data were smoothed using a 10 Hz lowpass Butterworth filter of a fourth order to create a linear envelope[35]. Moving average filter with a 500-millisecond window size was used on the 2-second MVIC EMG data, and the maximum of these processed MVIC EMG data was found across the two MVIC trials performed[36, 37]. The values of the SVC EMGs were calculated for each muscle and day of data collection following a similar procedure except that the mean values of the EMG data over the two seconds of SVC exertions were used instead of the 500-millisecond moving average filter. This was because during the SVC exertions the participant was exerting an acceptably steady torque (error of ±5%) while during the MVIC exertions, confirming a steady torque is not easily achievable due to the intense nature of the maximum exertions. The mean of the 2-second SVC EMG data was calculated for 20%, 40%, and 60% SVC EMGs, and were denoted as SVC-20%, SVC-40%, and SVC-60%.

**Data Analysis:**

JMP Pro 15 software package (SAS, Cary, NC) was used to perform a sequential data analysis including three steps:

***Step 1. The best predictor:*** To establish which %SVC value was the best predictor of the value of the MVIC on a given day, multiple linear regression technique was employed. All the seven possible models were evaluated that included predicting MVIC based on: 1) SVC-20%, 2) SVC-40%, 3) SVC-60%, 4) SVC-20% and SVC-40%, 5) SVC-20% and SVC-60%, 6) SVC-40% and SVC-60%, and 7) SVC-20%, SVC-40%, and SVC-60%. Both penalized-likelihood criteria of Akaike Information Criterion (AIC) and Bayesian Information Criterion (BIC) were employed as criterion to select the best SVC% EMG for each muscle.

***Step 2. Individual multipliers and predicted MVIC EMG:*** Once the best SVC% EMG predictor of MVIC EMG of the muscle was found, the multiplier that related the SVC% to the MVIC was found (per muscle and participant) for the Day 1 data. This multiplier was then applied to the appropriate SVC% value for the Day 2 data to predict the Day 2 MVIC per muscle and participant. Finally, this predicted MVIC EMG value of Day 2 was then compared with the actual MVIC EMG of Day 2 and the absolute percentage error was calculated for the muscle.

***Step 3. Between day reliability analyses:*** Finally, the reliability of the repeated peak generated moments, and MVIC EMG and SVC EMG values between days (Day1 and Day 2) were evaluated per muscle. Then, paired t-test or non-parametric Wilcoxon Signed-Rank test (if normality was violated using Shapiro-Wilk test) was used to evaluate if there was a significant difference between Day 1 and Day 2 for each desired variable (significance level of $p<.05$). zero.

## Results

***Step 1. The best predictor:*** The results of the multiple linear regression analyses revealed that the consistently effective model for predicting MVIC EMG for the biceps brachii, rectus femoris, and right and left neck flexor muscles utilized the SVC-60% values. These findings were consistent using both AIC and BIC criterion. Out of the 28 models (seven models × four muscles) there was only one model (left neck flexor) where combining SVC-20% and SVC-60% resulted in smaller AIC (0.9% decrease) and BIC (0.6% decrease) criterion compared to the model that only included SVC-60% (ranked second). For the right and left neck extensor muscles, the model that only included SVC-20% values led to the best regression based on both AIC and BIC criterion. Thus, for the sake of simplicity and easier usability, SVC-60% values were chosen for all studied muscles in calculating individual multipliers in the next step. For right and left extensor muscles the analysis was repeated for SVC-20% values and the results were compared to the model that employed SVC-60%.

***Step 2. Individual multipliers and predicted MVIC EMG:*** The mean (SD) of the ten individual multipliers for the six studied muscles were calculated (Table 1). The average absolute percentage error ranged from 45% (biceps brachii) to 26% (left neck extensor) (Table 2). Figures 3-4 compare the predicted and actual MVIC EMG of Day 2 for right and left neck extensor muscles using SVC-20% values and SVC-60% values. Ideally, all the data points in these figures would be on the line y=x. Furthermore, to put the average absolute percentage errors of this study in context, these values were compared to the average differences between the two muscle-specific MVIC EMG from the exertions performed on the same day (Table 3).

Table 1. The mean (SD) of the 10 individual multipliers that were used to predict MVIC EMG based on SVC-60%. The values in the square brackets for right and left neck extensor muscles allocate to the model based on SVC-20%.

| Muscle | Mean (SD) of the multipliers |
|---|---|
| Biceps brachii | 2.77 (0.98) |
| Rectus femoris | 2.49 (0.78) |
| Right neck flexor | 2.02 (0.58) |
| Left neck flexor | 2.40 (0.62) |
| Right neck extensor | 3.41 (1.90) [12.19 (5.19)] |
| Left neck extensor | 2.18 (0.58) [5.88 (1.59)] |

Table 2. The mean (SD) of absolute percentage error for the studied muscles based on SVC-60% values. The values in the square brackets for right and left neck extensor muscles allocate to the model based on SVC-20%.

| Muscle | absolute percentage error |
|---|---|
| Biceps brachii | 45 (41.8) |
| Rectus femoris | 27 (13.2) |
| Right neck flexor | 27 (13.4) |
| Left neck flexor | 33 (20.8) |
| Right neck extensor | 29 (24.7) [30 (32.8)] |
| Left neck extensor | 29 (25.4) [26 (16.1)] |

Table 3. The mean (SD) of absolute percentage difference between the two MVIC EMG exertions performed on the same day

| Muscle | absolute percentage difference |
|---|---|
| Biceps brachii | 20 (11.8) |
| Rectus femoris | 14 (12.1) |
| Right neck flexor | 13 (8.8) |
| Left neck flexor | 15 (12.5) |
| Right neck extensor | 20 (14.9) |
| Left neck extensor | 13 (10.0) |

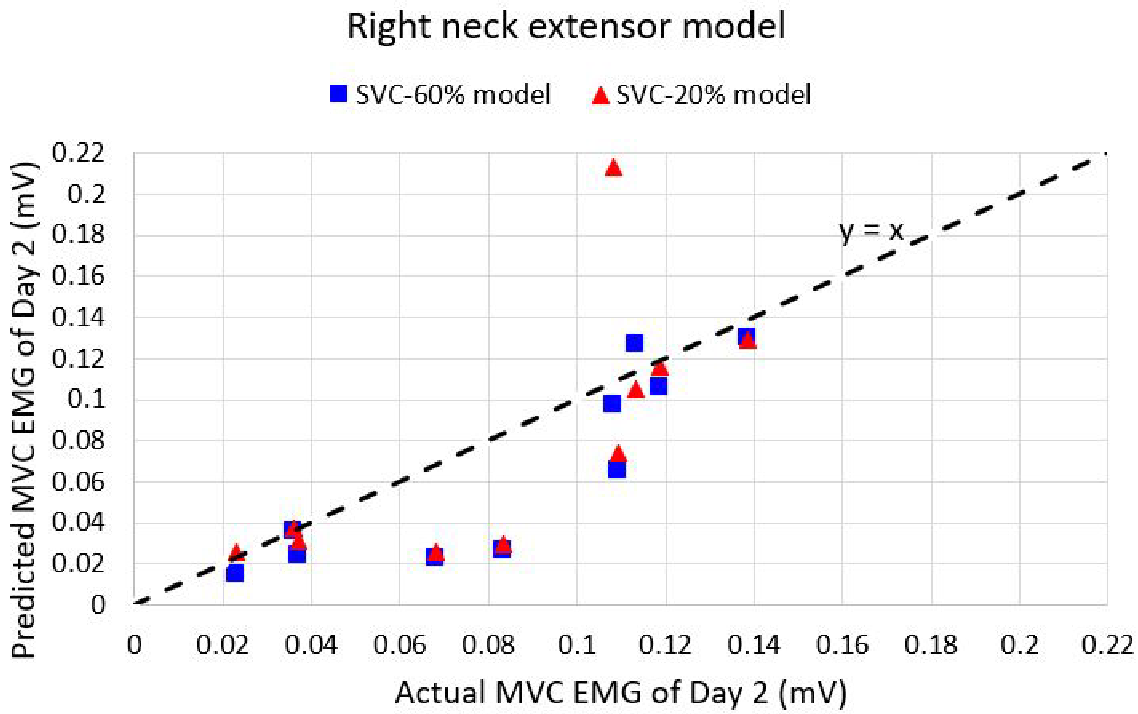



**Figure 3:** The predicted and actual MVIC EMG of Day 2 for right neck extensor muscle

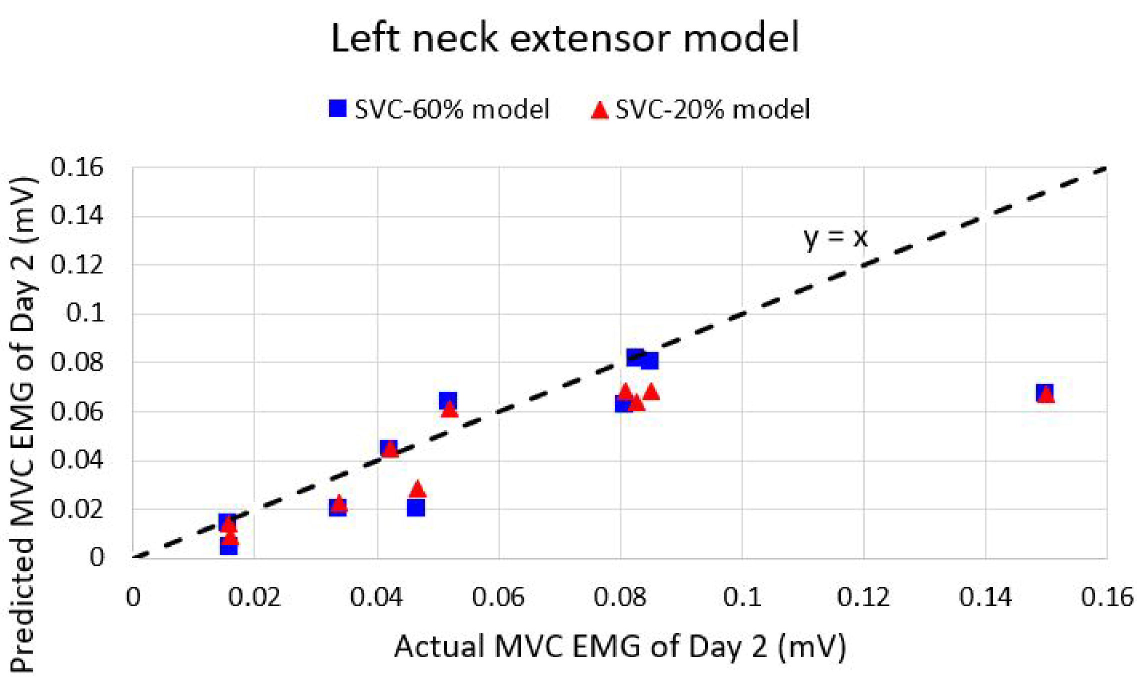



**Figure 4**: The predicted and actual MVIC EMG of Day 2 for left neck extensor muscle

***Step 3. Between day reliability analyses:*** The mean (SD) of the SVC and MVIC EMG values in millivolt and the peak generated moments in Newton-meter during Day 1 and Day 2 were calculated for each muscle (Table 4). The statistical analysis did not show a significant difference between Day 1 and Day 2 for any of these variables except for the peak generated moment of biceps brachii (p=.039). However, this increase was not reflected in comparing the corresponding average MVIC EMG values between Day 1 and Day 2.

Table 4. The mean (SD) of the SVC and MVIC EMG values in millivolt and the peak exerted moment in Newton-meter for Day 1 and Day 2 (Since the neck flexion and neck extension peak moment exertions were bilateral, the peak moments are only shown once.)

| Muscle | | SVC-20% | SVC-40% | SVC-60% | MVIC | Peak moment |
|---|---|---|---|---|---|---|
| Biceps brachii | Day 1 | 0.051 (0.026) | 0.139 (0.096) | 0.221 (0.102) | 0.585 (0.281) | **68.0* (17.3)** |
| | Day 2 | 0.044 (0.032) | 0.114 (0.073) | 0.188 (0.136) | 0.432 (0.318) | **73.7* (19.8)** |
| Rectus femoris | Day 1 | 0.015 (0.006) | 0.038 (0.052) | 0.048 (0.033) | 0.111 (0.055) | 193.4 (76.5) |
| | Day 2 | 0.012 (0.008) | 0.029 (0.013) | 0.048 (0.033) | 0.122 (0.067) | 200.3 (74.3) |
| Right neck flexor | Day 1 | 0.024 (0.015) | 0.061 (0.041) | 0.105 (0.064) | 0.201 (0.122) | 26.2† (11.3) |
| | Day 2 | 0.028 (0.017) | 0.063 (0.035) | 0.118 (0.067) | 0.220 (0.120) | 24.4† (11.8) |
| Left neck flexor | Day 1 | 0.023 (0.015) | 0.053 (0.031) | 0.108 (0.075) | 0.269 (0.217) | |
| | Day 2 | 0.029 (0.017) | 0.065 (0.042) | 0.120 (0.073) | 0.224 (0.123) | |
| Right neck extensor | Day 1 | 0.006 (0.003) | 0.014 (0.013) | 0.026 (0.024) | 0.071 (0.045) | 33.3† (12.2) |
| | Day 2 | 0.006 (0.003) | 0.012 (0.009) | 0.023 (0.023) | 0.084 (0.040) | 36.3† (13.4) |
| Left neck extensor | Day 1 | 0.008 (0.004) | 0.015 (0.011) | 0.025 (0.018) | 0.050 (0.029) | |
| | Day 2 | 0.008 (0.004) | 0.013 (0.008) | 0.022 (0.016) | 0.060 (0.041) | |

* Statistically significant difference between Day 1 and Day 2 ($p<.05$)

† One value for both right and left sides

## Discussion

The main purpose of this study was to develop and assess a method to predict the MVIC EMG of the muscle based on a system relating a muscle's SVC EMG to its MVIC EMG on a reference day. The motivation for developing this model was to decrease the risk of musculoskeletal injury that might result from multiple MVIC exertions across multiple days of experimental participation in vulnerable regions (e.g., neck) as well as joints with a history of injury. When considering specifically the results of the neck muscles, the average absolute percentage errors were more focused-in the range of 26% - 33%. These data would indicate that there is a trade-off between the required accuracy of the MVIC EMG and the risk of fatigue and injury due to exerting actual MVIC. Table 3 shows that the average absolute percentage difference between the two MVIC EMG exertions performed on the same day were between 13-20%, indicating that there is a natural variability in these MVIC EMG values.

Comparing the models for predicting MVIC EMG values (Table 2), it is inferred that SVC-60% is the consistently acceptable predictor of MVIC EMG values for all studied muscles. It is worth noting that SVC-80% may have resulted in better predictions of MVIC EMG. However, our pilot experiments revealed that exerting SVC-80% and holding it for two seconds is not easily achievable. Such exertions are very uncomfortable and may be prone to injury and discomfort. The proposed method for estimating MVIC EMG is intended to reduce the number of risky and uncomfortable exertions, thus SVC-80% exertions were not included in this study.

Using MVIC EMG data to normalize task EMG data is a well-established technique[22]. The principal benefit of normalizing to EMG obtained during an MVIC exertion is that it provides a clear standard that has physiological meaning relative to a participant's capacity/capability. For example, Sjøgaard and colleagues[38] explored the relationship between muscle blood flow and the magnitude of the muscle exertion, and they used %MVIC to establish the level of muscle exertion. Other methods have been introduced and studied such as normalizing with respect to the EMG collected during an SVC exertion[1, 12, 13] or normalizing task EMG relative to some EMG value that occurred during the task – normalized to the greatest value observed during the experimental task, for example. These methods do not require that the participants perform MVIC exertion. It is recognized however, that the interpretation of the resulting normalized EMG is a bit more limited. The method developed in the current study is based on actual MVIC EMG and would be suitable in studies that the participant is able to exert muscle MVIC, but they are required to attend on several days, or they are recruited for different studies. In these studies, muscle MVIC EMG could be collected once and be used to predict muscle MVIC EMG on other days.

It is clear that the accuracy and consistency of the magnitude of EMG data collected during MVIC exertions relies on the participant's motivation and willingness to provide true maximum voluntary contractions repeatedly[26, 27] and therefore limiting the number of MVIC exertions performed may limit the error that this might introduce. For example, in the current study, the experimental procedure was explained to the participants, and verbal encouragement was used during MVIC exertions. However, a simple analysis showed that the absolute percentage difference between the two MVIC EMG exertions performed on the same day (averaged across ten participants) was considerable (between 13-20%) indicating that there is a natural variability in these MVIC EMG values. Comparing this level of typical MVIC-MVIC variability to the average absolute percentage errors shown in our study (between 27%-33% for neck flexor/extensor muscles, 27% for rectus femoris, and 45% for biceps brachii; all SVC-60% models) would indicate that the predictive method developed, while not perfect in its predictions, is reasonably accurate considering the natural variability in MVIC exertions. The between day reliability analysis revealed that the average peak generated moment of biceps brachii increased significantly from Day 1 to Day 2, while the average MVIC EMG values for this muscle decreased (not significant) from Day 1 to Day 2 (Table 4). Additionally, the average absolute percentage error for biceps brachii (45%) was abnormally greater than the other studied muscles (27%-33%). These could be related to the experimental set-up during exerting the peak moments on the dynamometer (supplementary file 1). For the rectus femoris, neck flexion, and neck extension exertions, the participants were sitting with their trunk fixed to the dynamometer seating system (Figure 1b and 2). It could have limited their exertions strictly to the desired muscle/muscle group. However, the standing posture during the biceps brachii exertions (Figure 1a) may have led to employing additional muscles (e.g., leaning trunk backward slightly) despite all the instructions and experimenter's supervision.

There are a few limitations in this study that require attention before generalizing its results. Ten participants (one female) provided a limited and imbalanced sample size. It should be mentioned that the results for the female participant were found to be consistent with that of the nine male participants without revealing any specific trend or behavior. More studies with balanced and larger sample sizes will enhance the generalizability of our findings especially regarding the potential effects of gender. Only linear models were employed in this study to predict MVIC EMG based on SVC EMG. This decision was made based on the limited number of data points and to be consistent with the previous similar study[26] that assumed a linear relationship between EMG values and exerted moments. Nonlinear regression models with larger data sets and more levels of SVC EMG could potentially enhance the accuracy and reliability of predicted MVIC EMG values.

The main purpose of this study was to develop and assess a method to predict the MVIC EMG of the muscle based on a system relating a muscle's SVC EMG to its MVIC EMG on a reference day. We suggest that in each study individual multipliers should be calculated based on SVC-60% EMG values (per participant and muscle) to predict MVIC EMG values. Our results implied that there will be a trade-off between the required accuracy of the MVIC EMG and the risk of fatigue and injury due to exerting actual MVIC. Acceptability of the developed predictive technique depends on the study circumstances. For example, in a study that requires neck MVIC exertions on two different days with limited number of participants while a high accuracy in the EMG analysis is required, using MVIC EMG for normalization of the EMG data may be a better option compared to our predictive model. On the other hand, in another study where participants are required to attend several times, exerting neck MVIC EMG on each day of participation could question the feasibility of the study (e.g., risk of injury and IRB approval concerns). Additionally, in a study that investigates the muscle activity of surgeons during surgical procedures, exerting MVIC is not advisable because even a slight muscle cramp could have serious consequences for the surgeon and/or patient. Our developed predictive technique could be a better alternative in such studies where the participants are needed to exert MVIC only once and not necessarily on the day of data collection. The results of the EMG analysis would be affected by the approximations from our model; however, the findings of the study (as it is feasible to conduct the study using our model) could greatly recompense for these approximations.

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## References


1. De Luca CJ. The use of surface electromyography in biomechanics. *J. Appl. Biomech.* 1997;13(2):135-163. doi:10.1123/jab.13.2.135
2. Winkel J, Jorgensen K. Significance of skin temperature-changes in surface electromyography. *Eur J Appl Physiol Occup Physiol.* 1991;63(5):345-348. doi:10.1007/bf00364460
3. Koh TJ, Grabiner MD. Evaluation of methods to minimize cross-talk in surface electromyography. *J. Biomech*. 1993;26:151-157. doi:10.1016/0021-9290(93)90086-t
4. Mirka G. The quantification of EMG normalization error. *Ergonomics*. 1991;34(3):343-352. doi:10.1080/00140139108967318
5. Knutson LM, Soderberg GL, Ballantyne BT, et al. A study of various normalization procedures for within day electromyographic data. *J Electromyogr Kinesiol*. 1994;4(1):47-59. doi:10.1016/1050-6411(94)90026-4
6. Burden A, Bartlett R. Normalisation of EMG amplitude: an evaluation and comparison of old and new methods. *Med Eng Phys*. 1999;21(4):247-257. doi:10.1016/s1350-4533(99)00054-5
7. Sorbie GG, Grace FM, Gu YD, et al. Comparison of thoracic and lumbar erector spinae muscle activation before and after a golf practice session. *J. Appl. Biomech.* 2017;33(4):288-293. doi:10.1123/jab.2016-0209
8. Roth R, Donath L, Zahner L, et al. Muscle activation and performance during trunk strength testing in high-level female and male football players. *J. Appl. Biomech*. 2016;32(3):241-247. doi:10.1123/jab.2014-0303
9. Yaghoubi M, Esfehani MM, Hosseini HA, et al. Comparative electromyography analysis of the upper extremity between inexperienced and elite water polo players during an overhead shot. *J. Appl. Biomech*. 2015;31(2):79-87. doi:10.1123/jab.2014-0068
10. Marchetti PH, Uchida MC. Effects of the pullover exercise on the pectoralis major and latissimus dorsi muscles as evaluated by EMG. *J. Appl. Biomech*. 2011;27(4):380-384. doi:10.1123/jab.27.4.380
11. Juker D, McGill S, Kropf P. Quantitative intramuscular myoelectric activity of lumbar portions of psoas and the abdominal wall during cycling. *J. Appl. Biomech*. 1998;14(4):428-438. doi:10.1123/jab.14.4.428
12. Yang JF, Winter DA. Electromyography reliability in maximal and submaximal isometric contractions. *Arch Phys Med Rehabil*. 1983;64(9):417-420.
13. Burden A. How should we normalize electromyograms obtained from healthy participants? What we have learned from over 25 years of research. *J Electromyogr Kinesiol*. 2010;20(6):1023-1035. doi:10.1016/j.jelekin.2010.07.004
14. Allison GT, Marshall RN, Singer KP. EMG signal amplitude normalization technique in stretch-shortening cycle movements. *J Electromyogr Kinesiol*. 1993;3(4):236-244. doi:10.1016/1050-6411(93)90013-m
15. Suydam SM, Manal K, Buchanan TS. The advantages of normalizing electromyography to ballistic rather than isometric or isokinetic tasks. *J. Appl. Biomech*. 2017;33(3):189-196. doi:10.1123/jab.2016-0146
16. Monteleone BJ, Ronsky JL, Meeuwisse WH, et al. Lateral hop movement assesses ankle dynamics and muscle activity. *J. Appl. Biomech.*. 2012;28(2):215-221. doi:10.1123/jab.28.2.215

17. Tillman MD, Criss RM, Brunt D, et al. Landing constraints influence ground reaction forces and lower extremity EMG in female volleyball players. *J. Appl. Biomech*. 2004;20(1):38-50. doi:10.1123/jab.20.1.38
18. Perry J, Burnfield JM. Gait analysis: Normal and pathological function, Second Edition. *Gait Analysis: Normal and Pathological Function, Second Edition*. 2010:1-551.
19. Bolgla LA, Uhl TL. Reliability of electromyographic normalization methods for evaluating the hip musculature. *J Electromyogr Kinesiol.* 2007;17(1):102-111. doi:10.1016/j.jelekin.2005.11.007
20. Albertus-Kajee Y, Tucker R, Derman W, et al. Alternative methods of normalising EMG during cycling. *J Electromyogr Kinesiol*. 2010;20(6):1036-1043. doi:10.1016/j.jelekin.2010.07.011
21. Ball N, Scurr J. Electromyography normalization methods for high-velocity muscle actions: review and recommendations. *J. Appl. Biomech*. 2013;29(5):600-608. doi:10.1123/jab.29.5.600
22. Sinclair J, Taylor PJ, Hebron J, et al. The reliability of electromyographic normalization methods for cycling analyses. *J. Hum. Kinet.*. 2015;46(1):19-27. doi:10.1515/hukin-2015-0030
23. Dufour JS, Marras WS, Knapik GG. An EMG-assisted model calibration technique that does not require MVCs. *J Electromyogr Kinesiol*. 2013;23(3):608-613. doi:10.1016/j.jelekin.2013.01.013
24. McFarland DC, Brynildsen AG, Saul KR. Sensitivity of neuromechanical predictions to choice of glenohumeral stability modeling approach. *J. Appl. Biomech*. 2020;36(4):249-258. doi:10.1123/jab.2019-0088
25. Baggen RJ, van Dieen JH, Verschueren SM, et al. Differences in maximum voluntary excitation between isometric and dynamic contractions are age-dependent. *J. Appl. Biomech.* 2019;35(3):196-201. doi:10.1123/jab.2018-0215
26. Marras WS, Davis KG. A non-MVC EMG normalization technique for the trunk musculature: Part 1. Method development. *J Electromyogr Kinesiol*. 2001;11(1):1-9. doi:10.1016/s1050-6411(00)00039-0
27. McNair PJ, Depledge J, Brettkelly M, et al. Verbal encouragement: Effects on maximum effort voluntary muscle action. *Br. J. Sports Med.*. 1996;30(3):243-245. doi:10.1136/bjsm.30.3.243
28. Ettinger L, Weiss J, Shapiro M, et al. Normalization to maximal voluntary contraction is influenced by subacromial pain. *J. Appl. Biomech*. 2016;32(5):433-440. doi:10.1123/jab.2015-0185
29. Zellers JA, Parker S, Marmon A, et al. Muscle activation during maximum voluntary contraction and m-wave related in healthy but not in injured conditions: Implications when normalizing electromyography. *Clin Biomech*. 2019;69:104-108. doi:10.1016/j.clinbiomech.2019.07.007
30. Zeh J, Hansson T, Bigos S, et al. Isometric strength testing - recommendations based on a statistical-analysis of the procedure. *Spine*. 1986;11(1):43-46. doi:10.1097/00007632-198601000-00011
31. Battie MC, Bigos SJ, Fisher LD, et al. Isometric lifting strength as a predictor of industrial back pain reports. *Spine*. 1989;14(8):851-856. doi:10.1097/00007632-198908000-00014
32. Huelke DF, Nusholtz GS. Cervical-spine biomechanics - a review of the literature. *J. Orthop. Res*. 1986;4(2):232-245. doi:10.1002/jor.1100040212

33. Lindstroem R, Graven-Nielsen T, Falla D. Current pain and fear of pain contribute to reduced maximum voluntary contraction of neck muscles in patients with chronic neck pain. *Arch. Phys. Med. Rehabil.* 2012;93(11):2042-2048. doi:10.1016/j.apmr.2012.04.014
34. Hermens HJ, Freriks B, Disselhorst-Klug C. Development of recommendations for SEMG sensors and sensor placement procedures. *J Electromyogr Kinesiol*. 2000;10(5):361-374. doi:10.1016/s1050-6411(00)00027-4; http://seniam.org/
35. Gillette J C, Stephenson M L. Electromyographic Assessment of a Shoulder Support Exoskeleton During on-Site Job Tasks. *IISE Transactions on Occupational Ergonomics and Human Factors*. 2019; 7(3–4): 302-310.
36. Dahlqvist C, Enquist H, Löfqvist L, Nordander C. The effect of two types of maximal voluntary contraction and two electrode positions in field recordings of forearm extensor muscle activity during hotel room cleaning. Int J Occup Saf Ergon. 2020; 26(3):595-602. doi: 10.1080/10803548.2019.1599572. Epub 2019 Jun 21. PMID: 30932748.
37. Jensen C, Vasseljen Jr O, Westgaard RH. Estimating maximal EMG amplitude for the trapezius muscle: on the optimization of experimental procedure and electrode placement for improved reliability and increased signal amplitude. *J Electromyogr Kinesiol*. 1996;6(1):51-58. doi:10.1016/1050-6411(94)00012-3
38. Sjøgaard G, Savard G, Juel C. Muscle blood-flow during isometric activity and its relation to muscle fatigue. *Eur J Appl Physiol Occup Physiol*. 1988;57(3):327-335. doi:10.1007/bf00635992

## Supplementary File 1: Participant's position and the dynamometer set-up

### Biceps brachii

For the biceps brachii, the axis of rotation of the elbow joint was aligned with the axis of rotation of the dynamometer while the elbow joint was at 90º of flexion and the forearm was horizontal. The participant held the handle of the dynamometer and a wrist strap helped this grasping posture. An imaginary line between the rotation axis of the dynamometer and the handle of the dynamometer was parallel to the ground so that this line and the participant's forearm were in the same direction (Figure 1a in the main manuscript).

### Rectus femoris

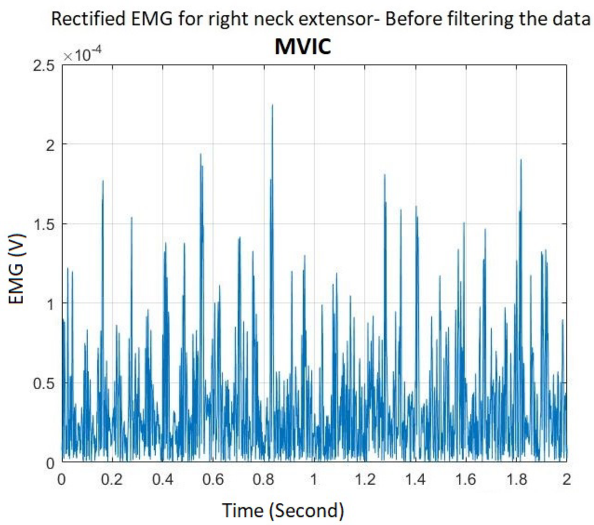
For rectus femoris, the participant sat on the dynamometer seating system, and their trunk was fixed to the back of the seating system using shoulder straps. The trunk and thighs made a 90º angle, and the thighs were parallel to the ground. The knee was at 90º of flexion so that their shin was perpendicular to the ground and the axis of rotation of the knee joint was aligned with the axis of rotation of the dynamometer (Figure 1b in the main manuscript).

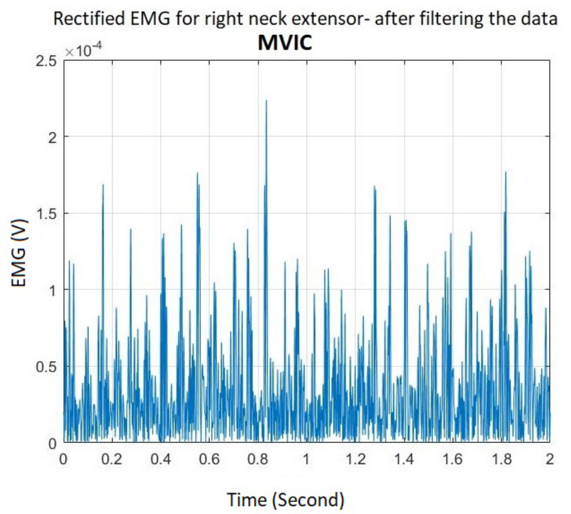
### Neck flexor and extensor muscles

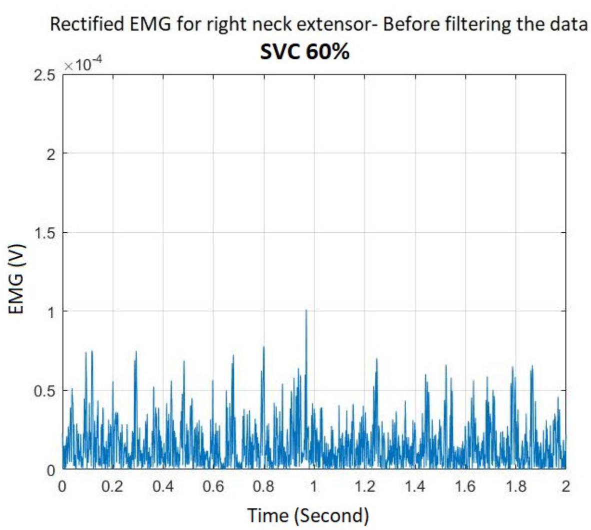
For neck flexor and extensor muscles, the participant sat on the dynamometer seating system, and their trunk was fixed to the back of the seating system using shoulder straps. Their trunk and thighs made a 90º angle, and their thighs were parallel to the ground. The handle arm of the dynamometer was adjusted to a comfortable height of the participant's forehead height (neck flexors) or posterior skull height (neck extensors) on the seating system while their neck was in an upright, neutral posture with no flexion, extension, lateral bending, or axial rotation. The height of the rotation axis of the dynamometer was adjusted at participant's seventh cervical vertebrae (C7) level when they were sitting on the dynamometer seating system.

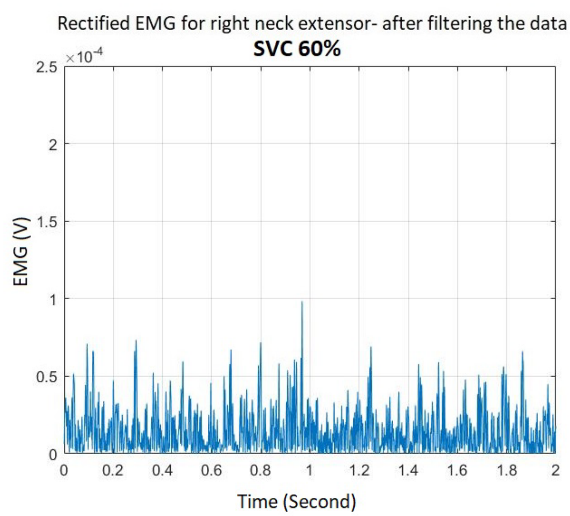
## Supplementary File 2: EMG data processing (filtering the data)

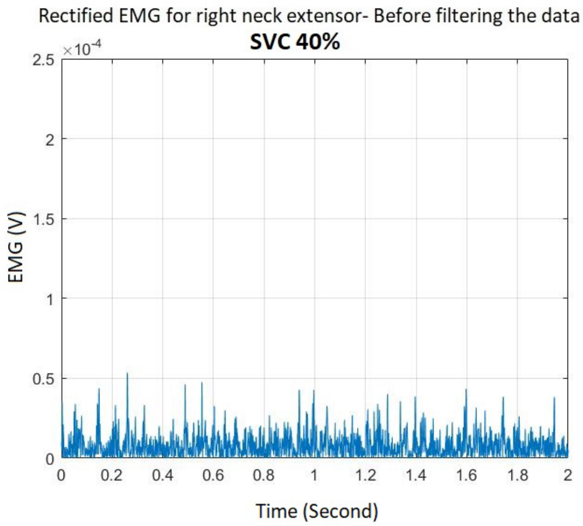
All EMG data were demeaned and filtered using a fourth order bandpass Butterworth filter (high pass=10Hz, low pass=400Hz) and a second order band-stop Butterworth filter (60 Hz and its closest harmonic alias (120 Hz)). The data for right neck extensor for one of our participants (Subject 3- Day 1) was chosen randomly and the graphs have been illustrated below.

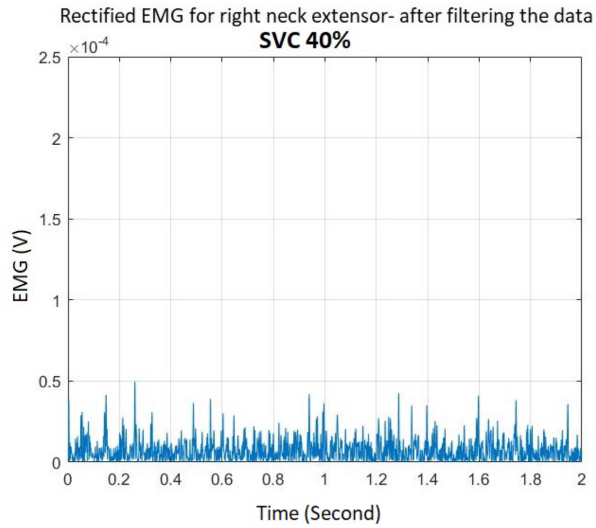


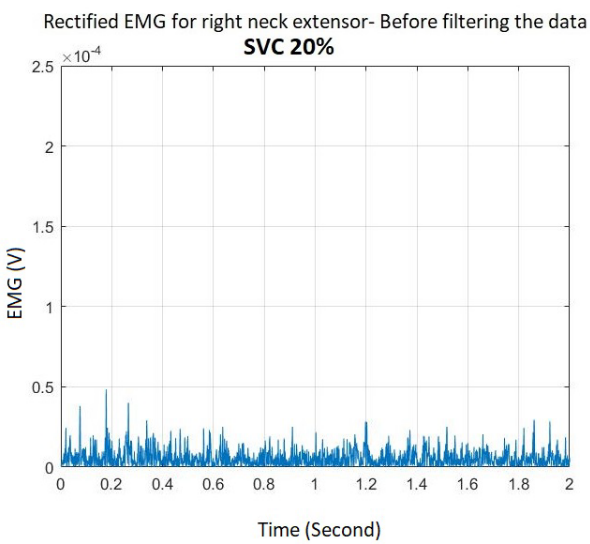



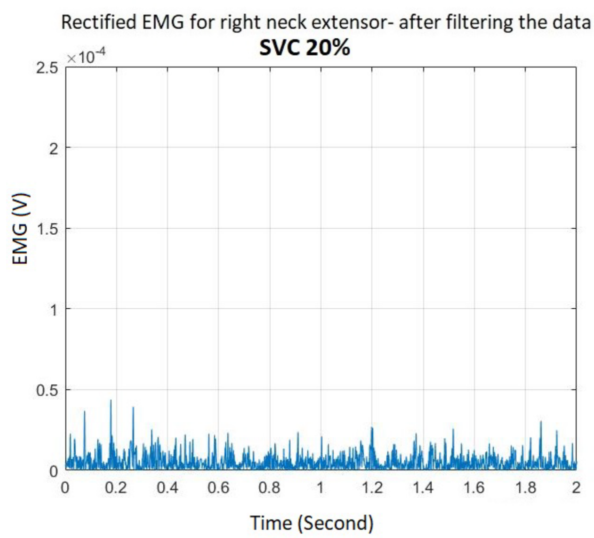




Rectified EMG for right neck extensor- Before filtering the data
SVC 40%
×10⁻⁴
EMG (V)
Time (Second)


Rectified EMG for right neck extensor- after filtering the data
SVC 40%
×10⁻⁴
EMG (V)
Time (Second)


Rectified EMG for right neck extensor- Before filtering the data
SVC 20%
×10⁻⁴
EMG (V)
Time (Second)


Rectified EMG for right neck extensor- after filtering the data
SVC 20%
×10⁻⁴
EMG (V)
Time (Second)