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The Advantages of Normalizing Electromyography to Ballistic Rather than Isometric or Isokinetic Tasks

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Isometric tasks have been a standard for electromyography (EMG) normalization stemming from anatomic and physiologic stability observed during contraction. Ballistic dynamic tasks have the benefit of eliciting maximum EMG signals for normalization, despite having the potential for greater signal variability. It is the purpose of this study to compare maximum voluntary isometric contraction (MVIC) to nonisometric tasks with increasing degrees of extrinsic variability, ie, joint range of motion, velocity, rate of contraction, etc., to determine if the ballistic tasks, which elicit larger peak EMG signals, are more reliable than the constrained MVIC. Fifteen subjects performed MVIC, isokinetic, maximum countermovement jump, and sprint tasks while EMG was collected from 9 muscles in the quadriceps, hamstrings, and lower leg. The results revealed the unconstrained ballistic tasks were more reliable compared to the constrained MVIC and isokinetic tasks for all triceps surae muscles. The EMG from sprinting was more reliable than the constrained cases for both the hamstrings and vasti. The most reliable EMG signals occurred when the body was permitted its natural, unconstrained motion. These results suggest that EMG is best normalized using ballistic tasks to provide the greatest within-subject reliability, which beneficially yield maximum EMG values.

Keywords: reliability, dynamic tasks, lower limb, sprinting, gait analysis

Electromyography (EMG) is widely used for identifying muscle function in patients with neurologic and orthopaedic pathologies. Studies have obtained EMG during minimally dynamic activities such as walking, stair climbing, or single hop tasks.[1–3] EMG from movements with increased velocities, or ballistic activities, have been used as predictive tools for anterior cruciate ligament (ACL) injury and hamstring strain.[4,5] In addition, ballistic activities have been performed on the ACL reconstructed population for rehabilitation and performance assessments.[6,7] The effectiveness of EMG as a tool for analyzing muscle activation across muscles, tasks, subjects, and testing sessions requires EMG signals to be normalized to a reference value.[8] The need for normalization becomes even greater between days if electrodes are removed and replaced without guide markings to ensure precise realignment. Signal magnitudes have been shown to vary by up to 31% due to electrode placement alone.[9] Even with consistent electrode placement, peak EMG differs between subjects and age populations.[10] It is important to examine the consistency of EMG signals if they are to be used as normalization values. More repeatable measures for normalization have been shown to improve sensitivity and intrasubject variability measures.[11]

A series of surface electrodes collecting EMG from various muscles identifies the contraction intensity used to perform an action. Large changes in EMG magnitude are also dependent on electrode placements due to anatomic traits, including: skin thickness, tissue impedance, distance from the skin's surface to the active motor units, and muscle fiber orientation relative to the surface EMG electrode.[9,12] Choosing the correct EMG normalization task is paramount in obtaining meaningful results, ie, comparing data across subjects, tasks, etc. Maximum voluntary isometric contraction (MVIC) values are a common normalization choice for studying EMG.[13] Because of the static nature of MVIC, anatomical factors that affect EMG, such as changing fiber pennation angles, joint angles, and joint velocities, are avoided.[12,14,15] In contrast, large changes in EMG magnitude during unconstrained kinematics have been studied[16] and rapid motions add even more variability to EMG signal collection. MVIC limits those variables while producing a maximal force, but certain muscles have been shown to elicit larger peak EMG signals during dynamic movements.[17] To date, a limited number of muscles have been compared for their reliability of peak EMG values.[18] The resulting EMG variability of various ballistic motions has not been thoroughly investigated and needs to be assessed.[19] Some studies suggest unconstrained motion may not influence variability in EMG. Repeatable EMG signals have been reported during several ballistic tasks, including: the lower leg muscles during running at a fixed speed[20] and rhythmic hopping,[21] the quadriceps muscles during maximum vertical jumping,[22] and both the quadriceps and hamstring muscles during a run/cut maneuver and drop jumps.[23] Repeatable values seen in a multitude of ballistic tasks indicate that anatomical factors may have a minimal influence, if any, on EMG variability.

Dynamic motion causes the distance from the electrode to the underlying motor units to change, which imposes signal variability.[24] Ballistic movements also require a different recruitment pattern than those of MVIC.[25] The possibility of changing recruitment patterns during motion may result in unreliable signals. Ballistic movements require rapid, maximal motor unit recruitment while distances from motor units to surface electrodes are changing. This may increase the likelihood of an optimum electrode to motor unit distance when the EMG signal is measured. In the case that this true maximum EMG is collected, the peak signal would not have the temporal range in which to fluctuate and would reduce

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EMG variability. The previously referenced studies that showed repeatable EMG signals during dynamic tasks all used familiar tasks such as running and jumping, which imply repeatable motion and consistent kinematics.[26,27] Repeatable kinematics, ie, reduced variability in motion, during high-level activities requires a consistent, rapid, maximum ballistic contraction which could lead to near-maximal EMG signals and increased reliability.

The purpose of this study is to assess the EMG magnitude and kinematic repeatability of dynamic tasks to determine peak EMG signal reliability. EMG signals from these dynamic movements (isokinetic, sprint and jump) will be compared to a constrained MVIC case. We hypothesize that the peak EMG during the ballistic tasks will be as repeatable as the MVIC, but with a greater magnitude. The single session, within-subject reliability will be comparable despite the anatomical changes during the dynamic movement because, though MVIC will maintain its constrained attributes, the ballistic cases require maximal exertion and dynamic control which will lead to a near maximal EMG signals, thus restricting variability.

Methods

Subjects

All subjects provided their informed written consent via a form approved by the institutional review board of the University of Delaware under the guidelines of the Declaration of Helsinki, prior to beginning. A total of 15 active (minimum of level 5 on the Tenger Activity Level Scale), uninjured, college-age subjects (average age of 24.1 ± 4.6 years, average height of 1.73 ± 0.10 m, and an average mass of 72.7 ± 13.5 kg) were recruited for study participation. Subjects were excluded from the study if they had any physical or neurological condition which would affect running or jumping. Subjects performed constrained MVIC and isokinetic (ISOK) tasks along with the dynamic tasks of maximum countermovement jumps and sprints.

Instrumentation and Setup

The Biodex 3 System (Biodex Medical Systems, Shirley, NY, USA) was used to fix joint angles and limit angular velocities during the MVIC and the ISOK tasks. EMG data were collected with a MA-300 EMG System (Motion Lab Systems, Baton Rouge, LA) for 9 leg muscles at 1000 Hz. The muscles included the medial and lateral vasti (VM, VL), rectus femoris (RF), biceps femoris (BF), semimembranosus (SM), medial and lateral gastrocnemius (MG, LG), soleus (Sol), and tibialis anterior (TA). Bipolar silver/silver chloride EMG surface electrodes (Myotronics, Inc., Kent, WA) were placed over these muscles. To ensure good conductivity, the electrode sites were shaved, lightly abraded, and cleansed with alcohol. Electrodes were held in place with elastic wraps. During the first task for each muscle group, signal amplifications were set and recorded on the EMG system to ensure a good signal-to-noise ratio and prevent saturation (voltage within ± 5 V).

An 8-camera, 3-dimensional motion capture system collected motion of retroreflective markers (Qualisys Motion Capture System, Gothenburg, Sweden) at 200 Hz. Retroreflective markers were placed on anatomical landmarks including bilateral acromions, iliac crests, greater trochanters, anterior and posterior iliac spines, femoral condyles, malleoli, and first and fifth metatarsals and C7 of the spine. Rigid marker clusters consisting of 4 retroreflective markers were subsequently placed on the lateral thighs, lateral shanks, and superior portion of the feet to establish a limb segment reference. Two additional reference markers were placed on the posterior of the feet and a cluster of 3 markers was placed on the sacrum. Anatomical landmark markers were subsequently removed, less the acromion and C7 markers, after a calibration trial. Kinematics of the knee and ankle joint angles were determined from the motion of the marker clusters during each of the ballistic movement trials. An AMTI BP6001200-2k force plate (Advanced Mechanical Technology, Inc., Watertown, MA) inlaid in the walkway measured ground reaction force (GRF) during the unconstrained dynamic trials, ie, sprinting and jumping.

Protocol

Following EMG electrode placement, subjects performed MVIC and ISOK trials on the Biodex 3 system. The subjects sat upright with their hips at 90° and their knees fully extended during the Biodex trials at the ankle. Each joint was secured to a fixed angle for the MVIC trials: the ankle joint fixed at 90° and knee joint at 60° of flexion. The subjects exerted maximal effort for 3 seconds of dorsiflexion, plantar flexion, knee flexion, and knee extension. The subjects performed ISOK tasks at 60°/s through their full range of motion (RoM) in both flexion and extension at each joint. The subjects pushed and pulled with their greatest effort resulting in a maximum flexion and extension in a single trial. The concentric phase demonstrated the greatest EMG magnitudes and EMG signals were only taken from the concentric portion of the ISOK tasks. The subjects rested at least 1 minute between trials to avoid fatigue, and each subject performed 3 trials of each MVIC and ISOK task. The subjects were given verbal encouragement to contract with maximal effort during every trial. The angles and velocities for the constrained cases were chosen because they demonstrated the greatest EMG amplitudes during pilot testing and are common in clinical assessment.

The unconstrained dynamic, or ballistic, trials were performed following the completion of the Biodex trials. The order of the ballistic tasks was randomized to avoid compounding fatigue effects between sprinting and jumping. A Brower timing system (Draper, UT) and a 20-m walkway with a floor-level force plate was used in the dynamic collections. The timing gates were set at the end of the force plate and 5 m prior. The sprint trial began with the subject at the back of the walkway in a standing lean, without contact with any surface besides the floor, ie, without starting blocks of any kind. The sprint task was performed at the subject's maximum running effort and a trial was accepted if it was within 5% of the subject's fastest trial and had a clean, single-foot landing on the force plate. Subjects repeated the task until a total of 5 acceptable trials were collected from each subject. The countermovement jump consisted of moving from a standing position into a squat, and then a maximal jump, using the arms to aid in reaching maximum take-off force and height, which was demonstrated for the subject prior to trials. Following demonstration, subjects practiced the movement until they felt comfortable. A trial was accepted if the vertical GRF, as recorded by the force plate, was within 5% of the other trials to ensure a maximal push-off was being elicited each time. Maximum jumping and sprinting trials were chosen since they were shown to have repeatable kinematics and require maximal effort.[27,28]

Signal Analysis

EMG and motion data were processed using Visual 3D (C-Motion Inc., Bethesda, MD) and were visually inspected for excessive external signal noise or signal dropout. The peak EMG for the sprint was taken from 2 gait cycles preceding toe-off from the force plate. The duration of 2 gait cycles was approximated due to the limited collection volume of the motion capture system. During the sprint, only the stance was consistently available. Using the measured stance time, gait time was determined by Novacheck and colleagues' reported gait cycle averages for sprinting of 36% stance, 64% swing.[29] Peak EMG for the countermovement jump was only measured during the push-off phase. Each muscle of each subject required a different amplifier setting based on the muscle's signal intensity during the initial trials. The amplification factor for each muscle group was standardized across subjects and tasks in order to compare EMG magnitudes. Additional details can be found in Appendix. EMG signals were processed by high-pass filtering using a fourth-order Butterworth filter with a 30-Hz cutoff frequency, which removed DC offsets, and then the signal was full-wave rectified. A linear envelope was created by low-pass filtering the signals at 4 Hz. The largest value of the linear envelope was the value used as the peak EMG signal for the trial. The GRF and motion data were low-pass filtered at 50 Hz and 6 Hz, respectively. The effect this processing had on the EMG signals is shown in Figure 1. Kinematics were only assessed during the stance phase of

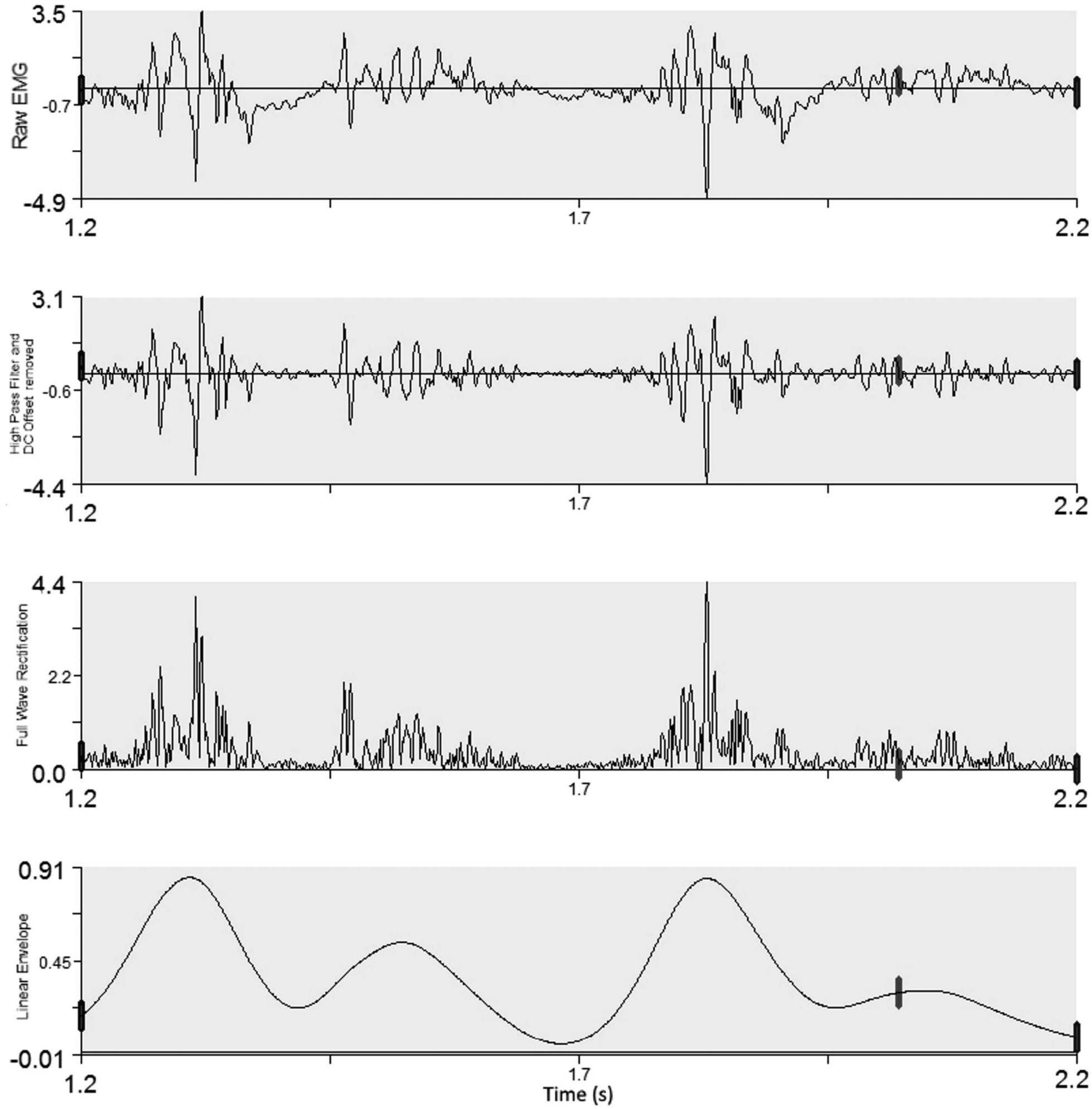

Figure 1 — The stages of filtering the electromyography (EMG) data from the rectus femoris during 2 complete gait cycles. Each graph shows the following steps: (top) raw EMG data, (top middle) high-pass filtering at 30 Hz and DC offset removed, (bottom middle) full wave rectification, and (bottom) low-pass filtering at 4 Hz to create a linear envelope. The grey hash marks represent the start of the 2 gait cycles, heel strike of the second gait cycle, and toe off (representing the end of the 2 gait cycles), respectively. Note the lack of soft tissue artifact after high-pass filtering.

Table 1 The range of motion (RoM) and velocity with the respective coefficient of variation (CV) averaged across subjects for each of the ballistic tasks

Joint	Ankle				Knee			
Measure (Unit)	RoM (°)	RoM (%CV)	Velocity (°/s)	Velocity (%CV)	RoM (°)	RoM (%CV)	Velocity (°/s)	Velocity (%CV)
Jump	75.0	2.7	538.5	5.0	102.8	3.0	373.0	1.7
Sprint	38.2	7.4	197.5	7.0	29.2	11.9	151.4	10.4

Note. The angular velocities and RoM measurements were collected during the stance phase of sprinting and the takeoff phase of jumping. All kinematic factors were repeatable with coefficient of variation <12%.

sprinting due to the limited collection volume of overground running.

Statistical Analysis

Analyses of the data were performed with the Statistical Package for the Social Sciences version 19 (IBM, Inc., Armonk, NY, USA). Intraclass correlation coefficients (3,1) (ICC) were computed for each muscle during each task within a single test session to establish within-subject reliability. An ICC greater than 0.75 was be considered reliable.[30] An ANOVA ($\alpha = .05$) was used to determine if a difference existed between the joint angles at which the peak EMG was developed during each task. An ANOVA ($\alpha = .05$) with a post-hoc analysis using a Tukey-Kramer honest significant difference (HSD) was used to compare the peak EMG values of the MVIC, ISOK, and the unconstrained, ballistic tasks to determine significant differences between each muscle independently. To demonstrate kinematic variability, the mean, within-subject coefficient of variation (CV) of the subjects' ankle and knee joint angles and velocity was used for the sprint and jump tasks.

Results

The mean number of jump and sprint tasks were 7 and 8 trials, respectively, to accomplish the 5 trials meeting the required criteria. A large difference existed between kinematic values of each dynamic task but these values remained consistent within subjects. The CV of the ballistic trials, reported as the average CV across subjects, resulted in consistent RoM and joint angular velocities for each joint, with the average CV ranging from 2.7% to 11.9% (Table 1). The mean RoM of the sprint during the stance phase was 29.2° at the knee and 38.2° at the ankle. The mean RoM for the jump was 102.8° at the knee and 75.0° at the ankle (Table 2). Both average joint velocities were greater during the jump than during the sprint (Table 2).

The triceps surae and vasti muscles had significantly greater EMG magnitudes during the ballistic tasks than the MVIC (Figure 2). However, the SM EMG signal during the ISOK task was significantly greater than both the MVIC and ballistic tasks. Each of the muscles' EMG during the dynamic and static cases were reliable, except the SM during the jump. The sprint task's peak EMG consistency was greater for 7 of the 9 muscles when compared to the MVIC and ISOK values (Table 2). The ICC values for the jump task were greater than those of the constrained motions for 5 of the 9 muscles. The ISOK case had the largest ICC values for 2 of the muscles and the MVIC, 1 (Table 2). There was no significant difference in the joint angle at which peak EMG signals were generated during the ISOK and jump tasks (Figure 3). The joint angle at which peak EMG was generated during the sprint task was unable to be evaluated due to the motion capture collection volume not being large enough to collect the entire RoM and the peak EMG occurring at different timings in the gait cycle.

Discussion

In this study, kinematic consistency and near-maximal EMG signals explained the single-session, within-subject reliability of peak EMG values during ballistic tasks and produced peak EMG consistency comparable to that of MVIC and ISOK tasks.

Table 2 The within-subject interclass correlation coefficients (confidence interval upper and lower bound) values for each muscle's EMG maximum

Muscle	MVIC	ISOK	Sprint	Jump
BF	**0.88** (0.74–0.95)	**0.94** (0.87–0.98)	**0.95** (0.90–0.98)	**0.89** (0.79–0.96)
LG	**0.85** (0.69–0.94)	**0.82** (0.61–0.93)	**0.96** (0.93–0.99)	**0.95** (0.89–0.98)
MG	**0.95** (0.89–0.98)	**0.87** (0.73–0.95)	**0.97** (0.93–0.99)	**0.96** (0.91–0.98)
RF	**0.92** (0.82–0.97)	**0.97** (0.92–0.99)	**0.92** (0.85–0.97)	**0.95** (0.90–0.98)
Sol	**0.82** (0.63–0.93)	**0.77** (0.55–0.91)	**0.93** (0.87–0.97)	**0.90** (0.81–0.96)
SM	**0.91** (0.81–0.97)	**0.88** (0.75–0.95)	**0.94** (0.88–0.98)	**0.69** (0.49–0.86)
TA	**0.99** (0.97–1.00)	**0.99** (0.98–1.00)	**0.96** (0.92–0.99)	**0.91** (0.83–0.96)
VL	**0.92** (0.83–0.97)	**0.86** (0.70–0.94)	**0.95** (0.91–0.98)	**0.95** (0.91–0.98)
VM	**0.90** (0.77–0.96)	**0.93** (0.84–0.97)	**0.94** (0.88–0.98)	**0.95** (0.91–0.98)

Abbreviations: BF = biceps femoris; LG = lateral gastrocnemius; MG = medial gastrocnemius; RF = rectus femoris; Sol = soleus; SM = semimembranosus; TA = tibialis anterior; VL = lateral vasti; VM = medial vasti.

Note. The sprint trial's EMG signals were the most reliable and the jump trial was the only task with an unreliable signal (SM).

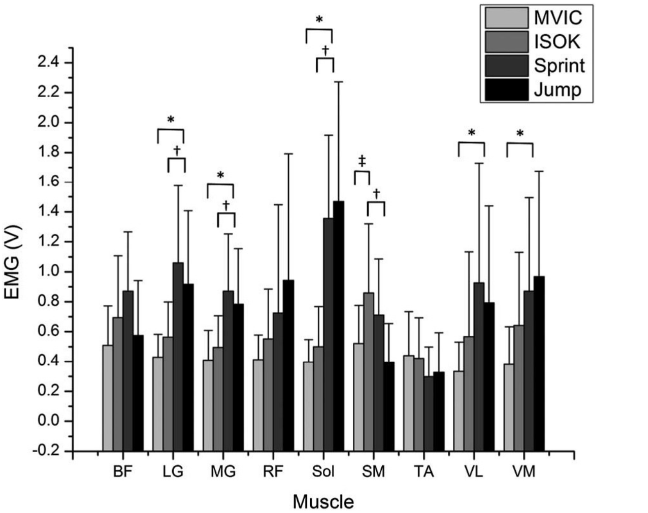

Figure 2 — The average peak (standard deviation) electromyography (EMG) signal magnitude (V) for each muscle across all subjects during each task: maximum voluntary isometric contraction (MVIC), isokinetic contraction (ISOK), sprint, and jump. The (*) and (‡) denote a significant difference ($p < .05$) in magnitudes between the ballistic and the MVIC case and the ISOK and the MVIC case, respectively. The (†) indicates a significant difference ($p < .05$) between the ballistic and ISOK peak EMG for each muscle during the tasks. Note that the ballistic cases were significantly greater for 5 of the 9 muscles including all the quadriceps and triceps surae muscles. BF = biceps femoris; LG = lateral gastrocnemius; MG = medial gastrocnemius; RF = rectus femoris; Sol = Soleus; SM = semimembranosus; TA = tibialis anterior; VL = lateral vasti; VM = medial vasti.

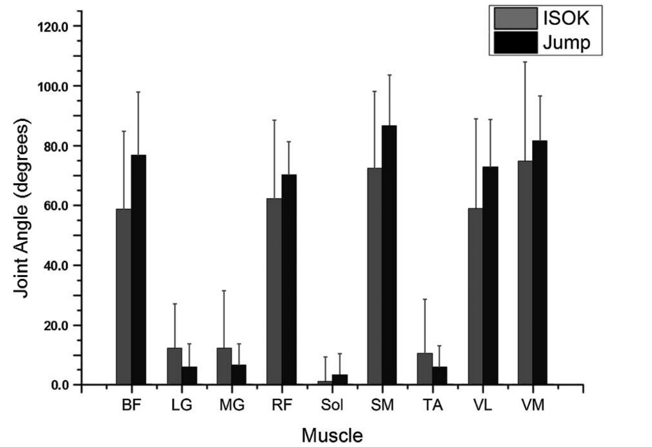

Figure 3 — The average peak (standard deviation) joint angle (degrees) across all subjects at which the peak electromyography (EMG) signal was generated for each of the muscles during the isokinetic contraction (ISOK) and jump tasks. Note the lack of significant differences between the ISOK and jump tasks along with the large standard deviations for each muscle indicating no consistent angle at which the maximal EMG signal was generated between subjects. BF = biceps femoris; LG = lateral gastrocnemius; MG = medial gastrocnemius; RF = rectus femoris; Sol = Soleus; SM = semimembranosus; TA = tibialis anterior; VL = lateral vasti; VM = medial vasti. The joint angles represent knee flexion angle for the BF, RF, SM, VL, and VM, plantar flexion for the LG, MG, and Sol, and dorsiflexion for the TA.

Our hypothesis was supported as we found that the EMGs from ballistic tasks were as repeatable as those of the constrained tasks (Table 2). Peak EMG is reliable for all muscles in both the MVIC and ISOK tasks, as demonstrated in previous studies.[8,31] In this study, all of the muscles measured during the sprint and 8 of the 9 muscles measured during the countermovement jump produced repeatable peak EMG values. The within-subject reliability of the EMG during the sprint and jump tasks in this study contrasts with that of previous reports that found ballistic EMG values to be less reliable than those of the MVIC.[21–23] The differences between these results may be explained by methodology. This study used peak EMG during maximal-effort ballistic tasks, whereas previous studies evaluated integrated EMG signals from dynamic movements, such as hopping, which may not have required maximum contractions.[21,22] In addition, kinematics were not reported in the referenced studies, so the repeatability of the motions are unknown.

MVIC constrains kinematics to limit anatomical changes during contraction and produces reliable EMG signals. Although there were large differences in magnitude, joint RoM, and velocity between ballistic tasks, the peak signal variability within subjects remained low. We attribute this to the kinematic consistency within subjects. Increased RoM causes relative location changes between the muscle, and therefore underlying motor units, and the surface electrodes. This leads to signal collection across a changing area of the muscle and varying electrode distances from multiple motor units. However, with the RoM being repeatable (CV < 12%), the area of the muscle fibers and relative location motor unit to the electrode was also repeated, leading to reliable EMG signals. Similarly, fluctuations in EMG signal magnitude have also been associated with changes in joint velocity,[15] but the repeatable joint velocities (CV < 10.5%) at the subjects' maximal effort requires a repeated contraction intensity and recruitment strategy. Our data are consistent with previous running gait findings yielding a within-day average knee flexion CV of 10.7%[20,26] and jumping data that have reported consistent knee and ankle kinematics during dynamic tasks.[27]

Our data revealed that MVIC does not produce the largest EMG signals, which is in accordance with previous research.[17,18,20] The rapid recruitment of motor units in concert with maximum effort, large joint velocities, and increased RoM during the ballistic tasks are the most likely mechanisms that influenced the increased ballistic EMG values compared to the MVIC (Figure 1, Table 2). The joint RoM and velocities impact the peak EMG through the changing EMG collection site within the detectable region. As the muscle contracts, the motor unit moves relative to the skin and EMG electrode.[12] This is applicable to any contraction, static or dynamic. However, the dynamic motion has the additional benefit of muscle shortening by both the muscle contraction and the changing joint angles. This large range of distances between the motor units and the electrode expands the opportunities to measure EMG from optimal locations when trying to obtain the maximal signal. Within this logic, increased joint RoM would lead to measuring EMG from many joint angles and finding the maximal EMG signal. However, when comparing magnitudes in this study, peak EMG was not found during the ISOK being performed at full RoM and therefore other factors must be taken into account. The range of motion not being the only variable in this study leading to a maximal signal was further supported by the fact that there was no consistent angle at which peak joint forces were produced during the ISOK and jump trials. The lack of significance and large standard deviations shown in Figure 3 imply that each subject for this trial had a different joint angle at which their maximum EMG was produced for each movement. This leads to the idea that increased EMG must also be a result of altered motor recruitment strategies used during ballistic movements[32]; thus it depends not only the dynamic nature of the contraction but the control demanded. The unconstrained nature of the ballistic movement forces the neuromuscular system to make "decisions" based on the demands of the task and external factors. When the goal of the system is maximum force output and rate of force production, the neuromuscular system appears to send greater and more consistent signals to achieve the demands.

It is important to note that the muscles with significantly increased signals during the ballistic tasks were the propulsive muscles for those tasks (VL, VM, MG, LG, and Sol). The quadriceps and plantar flexors are the primary contributors to the acceleration of the center of mass during gait[33] and these muscles are also the primary contributors to maximal-jump heights.[34] The significantly larger peak EMG signals of these muscles were products of rapid changes in joint angles in order to generate the maximum sprint speeds and jump heights induced by intense muscle recruitment. These tasks were also unconstrained, which means they required coordination and recruitment in a way different than those of a constrained dynamometer task.

The peak EMG value of the hamstrings, which operate primarily as knee flexors and hip extensors, was not significantly increased between the static and ballistic contractions. In contrast, the ISOK had a significantly greater SM peak EMG signal than the other tasks (Figure 2). The mean peak EMG of the hamstrings during the sprint and jump were larger than that of the MVIC task for the BF, but the large standard deviations of the BF EMG signals contributed to the lack of significant difference between the ballistic tasks and the MVIC task. Some of the variability in the sprint trials may be attributed to varying levels of subject training and technique. Novice runners use the hamstrings as an eccentric, momentum-reducing muscle during the swing phase; whereas highly-trained, elite runners use the hamstrings concentrically to aid in propulsion during stance. In contrast, both novice and elite runners use the quadriceps and triceps surae as propelling muscles.[29] During the jump task, the hamstrings do not play a large role in vertically propelling the center of mass. The propulsive strategy and recruitment of the knee flexors and hip extensors of each subject varied during the jump between these nonelite athletes, causing the amount of hamstring recruitment to deviate. Therefore, it was not surprising that the lowest EMG repeatability was found for the SM during the jump.

The nonelite status of the participants also created the need for ballistic task acceptance criteria, ie, trials within 5% of the speed (sprint) and jump height. The muscle coordination and neural control strategies of nonelite athletes are not perfectly tuned, and hence the possibility of different outcomes exists for each attempt, despite perceived maximal effort. Acceptance criteria limit the coordination variation by ensuring that the results are similar. However, even though there were required criteria to be met, the jumping task only required an average of 2 additional trials and sprinting required 3 additional trials. Sprinting retrials were also subject to proper landing on the force plate, which accounted for an average of 1 additional trial per subject.

The muscle groups that had significant EMG magnitude increases during the ballistic tasks also had the highest ICC. Although superficially this may suggest that magnitude and reliability are related, we believe this is not the case. Karamanidis and colleagues found no distinct correlation pattern of ICC to maximum EMG in relation to effort as determined by increasing running velocity with no increase in relative reliability.[20] We found a similar trend, as the ISOK had increased magnitudes for the LG, MG, Sol, SM, and VL compared to the MVIC case, but the ICC values of the EMG for these muscles during ISOK tasks were lower than the MVIC ICC values (Figure 1, Table 2). This again suggests control demands of unconstrained dynamic tasks on the neuromuscular system creates a peak output to reach maximum effort and this causes more reliable peak EMG values, since it has reached an EMG ceiling and is unable to increase its neural output.

The number of MVIC trials from which single session, within-subject reliability has been determined varies from study to study. Some studies collect more than 3 trials in order for the EMG reliability to be obtained,[17] while others only provide 3 trials due to the fatiguing effects of static trials.[11,23] This study was especially sensitive to fatiguing effects of MVIC due to the number of ballistic activities following the MVIC tasks. Pilot data in this study showed EMG being consistent after only 3 trials, along with research of other athletic-based research studies presented on EMG, and therefore 3 trials were chosen for both the static and isokinetic trials.

Certain limitations existed within this study. Surface electrodes were used to collect dynamic EMG due to the complications of electrode movement within a muscle belly associated with the use of fine wire electrodes during ballistic activity. This use of surface electrodes leads to the possibility of cross-talk between some muscles. The RF has the potential to collect signals from the large signal-generating vastus intermedius. The Sol has the potential to collect signals from the more superficial gastrocnemii. To avoid this error, the electrode for the Sol was placed distal to the MG and LG. The distal portion is not the largest part of the muscle but the signal available at that location is a strong indicator of Sol activation. In addition to these limitations on electrode placement, the RoM and joint velocity at the knee for the sprint case were limited by the collection volume within the lab environment. The collection was taken overground and the sprint strides were sometimes greater than the 3D motion capture volume, preventing the collection of an entire gait cycle, and therefore the values reported are based on a single stance time and cycle time based on normative averages. This also hindered the ability to evaluate the joint angle at which each muscle produced its maximum signal.

Conclusion

The results from this study support our hypotheses that the ballistic trials were as reliable as the MVIC case, and with greater EMG magnitudes. EMG magnitude and variability is a product of many factors, but repeatability can be maintained if the subjects are able to repeat joint kinematics and contraction intensity, reaching near-maximal EMG values. It is possible that the kinematic consistency, not the predefining or prevention of motion, is the key aspect of the constrained tasks producing reliable EMG signals. The unconstrained cases producing reliable values present a strong argument for allowing kinematic freedom during tasks to produce reliable peak EMG signals so long as the tasks themselves are repeatable by the subject population. Though these data are directly applicable to subject populations with a high level of mobility and function, such as athletes or injured populations near full recovery, inferences can be made to support the use of ballistic tasks with repeatable kinematics to yield reliable EMG normalization values.

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Appendix

Each of the MA-300 EMG system receivers (Motion Lab Systems, Baton Rouge, LA) have individual amplifiers for each EMG signal. Amplifications are raised or lowered to ensure the best signal to noise ratio without the signal exceeding the ±5 voltage limit. During the collections, each of the amplifications are optimized for each of the subject's muscle and amplification values are collected. The nature of EMG amplification for proper signal to noise ratios leads to an inability to compare EMG magnitudes across muscles or subjects. A global amplification value is required to facilitate a comparison of EMG magnitudes. This global amplification was selected as the most common value for each muscle, across subjects. An example of ensuring an EMG signal was amplified by the global amplification value starts with an EMG signal with an amplification of 5700. In this example, the global amplification for that muscle across subjects is 4000. The entire, raw EMG signal is multiplied by 4000 and divided by 5700. The signal has now been amplified by the same value as the rest of the EMG signals for which it is being compared.