Effect of a hip exotendon system on lower body muscle activity during forward skating accelerations

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Abstract

This study investigated the activity of six lower body muscles during a forward skating acceleration task from a stationary starting position while wearing a novel exotendon system designed to assist hip extension. Three levels of tension band stiffnesses were assessed (soft, medium, stiff) along with a control condition. Measures of root mean square (RMS) electromyographic (EMG) muscle activity were analyzed from five elite male ice hockey players and normalized as a percent of dynamic maximal activity for each respective muscle. Mixed ANOVAs compared band conditions across all stance phases captured during the task for each muscle, and principal component analyses were run for two hip extensor muscles during an 'acceleration' stride and a 'steady state' stride. The exotendon band conditions revealed no effect on muscle activity regarding both RMS EMG measures and principal component scores. Stance events showed significant differences in activation of the muscles over the length of the acceleration task, both supporting and linking previous ice skating literature. Despite the absence of change being driven by the exotendon system, the findings indicate that the design may need more extreme band stiffness properties and longer skating tasks to elicit muscle activity changes at the hip joint. Future work should be done to investigate this type of equipment in a larger variety of hockey skating scenarios such as submaximal efforts, full stride skating, and to re-assess the system using more resistive elastic elements.

Abrégé

Cette étude a examiné l’activité de six muscles inférieurs du corps lors d’une tâche d’accélération de départ-arrêté de hockey sur glace ; en utilisant un système exotendon conçu pour faciliter l’extension de la hanche. Trois niveaux de rigidité de bande ont été évalués (souple, moyen, rigide) ainsi qu'une condition de contrôle. Les mesures de l'activité musculaire électromyographique (EMG) ont été analysées à partir de cinq joueurs élite de hockey sur glace masculins et normalisées sous forme de pourcentage d'activité maximale dynamique pour chaque muscle respectif. Des ANOVA mixtes ont comparé les conditions de bande pour toutes les événements de poussées capturées durant la tâche pour chaque muscle, et des analyses en composantes principales ont été effectuées pour deux muscles extenseurs de la hanche lors d’une foulée « accélérée » et d’une foulée « glissé ». Les conditions de bandes d'exotendon n'ont révélé aucun effet sur l'activité musculaire en ce qui concerne les mesures EMG et les scores en composantes principales. Il a été constaté que les événements de poussée montraient des différences significatives dans l'activation des muscles au cours de la durée de la tâche d'accélération, soutenant et reliant les connaissances précédentes de patinage sur glace. Les résultats indiquent que la conception peut nécessiter des rigidités de bande plus extrêmes pour provoquer des changements au niveau de l'articulation de la hanche lors d’une tâche de patinage à effort maximal. Des travaux futurs devraient être menés pour étudier ce type d’équipement dans une plus grande variété de scénarios de patinage, tels que les efforts sous-maximaux, des tâches de patinage plus long et pour réévaluer le système en utilisant des éléments élastiques plus résistifs.

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Contribution of Authors

Michael Solomon, the MSc candidate, Department of Kinesiology and Physical Education, McGill University, was responsible for the processing, analysis and writing of this thesis. Alongside these responsibilities, the candidate helped with the experimental setup and collection of the data while completing all requirements associated with completion of a master's thesis at McGill University. A number of other individuals had a role in the completion of this study. David J. Pearsall, PhD, Associate Professor, Department of Kinesiology and Physical Education, McGill University, the Candidate's supervisor possessed an essential advising role for this thesis, especially in regard to analysis and result interpellation. Shawn Robbins, PhD, Assistant Professor, School of Physical and Occupational Therapy, McGill University helped develop the pipelines for the analysis of the EMG data and advised in the decision-making process for the statistical analysis of the data.

Phillippe Renaud, MSc, Department of Kinesiology and Physical Education, McGill University, provided assistance throughout the project's entirety with the research design, data collection, and subsequent data analysis as the lab research assistant. Kristie Liu, MSc, and Neil MacInnis, MSc candidate, Department of Kinesiology and Physical Education, McGill University, assisted with the data collection. Brian McPhee, MSc, developed the exotendon system used to aid skating, conceptualized the experimental design, allowed for the simultaneous recording of extraneous surface electromyography (EMG) to occur during his protocol and assisted with data collection. Aiden Hallihan, MSc, contributed to the experimental design with the choice of muscles recorded in accordance to his skating stop-and-go study task, and assistance with data collection.

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Chapter 1: Introduction and Literature Review

1.1 Introduction

Innovations in sports often focus on augmenting athletic ability through changes in sporting equipment. While much of the ice hockey industry has focused on skate and stick development to achieve enhancements in performance, alterations in worn protective equipment remains unexplored to this end. The concept of exoskeletons and/or exotendons to aid human movement have gained momentum with two design streams; active and passive. Passive designs specifically are characterized by the employment of unpowered elements to store and release elastic energy in unison with the human body. These exo-systems generally come with an associated low weight cost and as such possess potential for integration into sporting equipment. To date, numerous movement studies have shown reductions in metabolic activity and net gains in power output with use of passive exoskeletons (Collins, Wiggin, & Sawicki, 2015; Elliott, Sawicki, Marecki, & Herr, 2013; Grabowski & Herr, 2009; Lee et al., 2017). Select discussions suggest the benefits of such equipment may be in relation to alterations in joint stiffness regulation during human movement imposed by the parallel acting assistive elements (Y. H. Chang, Roiz, & Auyang, 2008; Farley & González, 1996; Ferris, Bohra, Lukos, & Kinnaird, 2005; Grabowski & Herr, 2009). To note was the variation that has been found in individual response towards performance enhancing elements (Wannop, Worobets, Madden, & Stefanyshyn, 2016). The exact mechanisms behind exo-system alterations to human movement remain undefined, with some suggestions leaning towards changes in muscle activation characteristics such as decreases in muscle activity associating with lowered metabolic cost and power as well as changes in activation patterns or structures (Collins et al., 2015; Farris, Robertson, & Sawicki, 2013; Ferris, Bohra, Lukos, & Kinnaird, 2005; Steele, Jackson, Shuman, & Collins, 2017).

In the ice hockey biomechanics field, recent studies have edified the kinematic understanding of a multitude of skating movements through the employment of modern 3-D motion capture technology on ice (Buckeridge, LeVangie, Stetter, Nigg, & Nigg, 2015; Hallihan, 2018; McPhee, 2018; Renaud et al., 2017; Shell et al., 2017). Some of these studies have additionally taken to utilizing electromyography (EMG) as means of strengthening the comprehension of skating performance (Buckeridge et al., 2015; Hallihan, 2018). These investigations capturing maximal forward skating acceleration starts of ice hockey players have identified crucial movement kinematics and associated muscle activity for dictating peak task execution. Explosive, running-like strides have been found to dictate overall skating speed with the skater's performance ceiling set by their plyometric strength and motion limitations imposed by equipment (Robert-Lachaine, Turcotte, Dixon, & Pearsall, 2012). The aforementioned findings have led to the creation of a low profile passive exotendon system designed to improve linear skating performance through implementation of tension bands across the hip joint (McPhee, 2018). Acceleration task completion times were previously found to improve with employment of individualized band stiffness, yet kinematics revealed little difference in movement execution. As such, this study is looking to investigate the activity of six superficial lower limb muscles to better deduce the mechanisms behind the passive exotendons performance altering effects.

1.2 Review of Literature

This section of the paper will provide a summary of previous research completed in skating kinematics. A brief introduction of hockey's evolution as well as skating fundamentals will be discussed. An overview will be presented of kinematic and electromyography (EMG) collection techniques in sport research with primary focus on ice skating sports. Subsequently, a review of exoskeleton/exotendon implications for augmenting locomotion performance will be presented. Studies of lower body muscle activity while employing exo-systems will conclude this section. This will set the stage for the study's rationale, objectives, and hypotheses.

1.2.1 History of Ice Hockey

In terms of locomotion, the human species has evolved to maximize our movement efficacy in relation to the desired outcome. For example, changes in locomotion at the earliest stages of human evolution saw ancestral chimpanzees' transition from quadrupedal walking to the now retained form of bipedal walking as result of a notable decrease in metabolic cost (Sockol, Raichlen, & Pontzer, 2007). Jump to modern history, humans have continually found better methods of transportation through novel technology, often in response to the surrounding environment. As humans moved into northern Europe they faced persistent snow and ice which made basic bipedal walking inefficient and sometimes unfeasible as a method of locomotion when traversing long distances. In response to this obstacle over 3000 years ago, these humans made the first prototype skis and skate tools (Formenti & Minetti, 2007).

Initial skates were extremely primitive in design (Formenti & Minetti, 2007). Over the next two millennia, skates gradually transformed to improve functionality and optimize locomotion. Ice skating as a competitive sport commenced between the 16th and 17th century where athletes strapped on rudimentary wooden skates instrumented with iron runners to race one another (De Koning, Houdijk, De Groot, & Bobbert, 2000). It was not until the 1880's that the sport of ice hockey was believed to have been conceived in Windsor, Nova Scotia with roots drawing from other stick-ball games of the time (Pearsall, Turcotte, & Murphy, 2000).

The modern sport of ice hockey is enjoyed both recreationally and competitively by children and adults, men and women on natural outdoor ice surfaces to indoor synthetic ice environments. Concurrent advances in ice hockey include improved equipment and infrastructure, as well as greater participant development including coaching, game strategy, training, and injury prevention (Pearsall et al., 2000).

1.2.2 Fundamentals of Hockey Skating

Formenti and Minetti (2007) describe ice skating as human-powered locomotion aided by a blade or blade-like structure between the foot and ice surface that allows individuals to glide with great efficiency. In the sport of ice hockey, players use hockey skates to move with agility and speed (Pearsall et al., 2000). Gliding blades on an ice surface (between 0 and -10°C) have a low coefficient of friction, thus provide little resistance to gliding; however, blades do not provide the needed push-off resistance for forward acceleration due to backward leg extension such as in running. Instead, propulsion reaction forces must be generated perpendicular to the blade and oblique to the body's line of progression. Furthermore, as the skate blade glides forward it must create a shallow groove (2 to 5 mm deep) into the ice surface to create a transient grip edge for sufficient perpendicular blade friction for push-off. This relationship makes skating a unique form of locomotion for humans (de Koning, de Groot, & Ingen Schenau, 1991; de

Koning, Thomas, Berger, Groot, & Van Ingen Schenau, 1995; Mark Denny, 2011; Pearsall et al., 2000).

Skating is the fundamental skill in ice hockey. The sport is categorized as a multi-player, tactical invasion game. Players can spend substantial time on the ice during a 60 minute game through numerous short power shifts (45 sec). A majority of maneuvers rely upon some form of the forward skating movement pattern (Upjohn, Turcotte, Pearsall, & Loh, 2008). The forward skating stride itself, whether in ice hockey or in speed skating, can be classified as a biphasic human motion with a support and swing phase (de Koning et al., 1991; Marino, 1977; Pearsall, Turcotte, Lefebvre, & H, 2001; Pearsall, Turcotte, Levangie, & Forget, 2013; Upjohn et al., 2008). The stride commences at initial skate blade ice contact, followed by three defined phases: glide, push-off, and recovery (de Koning et al., 1991; Pearsall et al., 2000). Glide is defined as the phase in which no propulsive force is being produced and the player glides parallel to the blade's path. The subsequent push-off phase generates the strides propulsion as the blade first externally rotates then is followed up by a rapid extension of the hip and knee coupled with ankle planter flexion until the leg reaches a state of full extension with the blade leaving the ice surface. The final phase of the stride cycle, the recovery phase, successively takes place as the now fully extended, non-weightbearing limb is flexed and swung forward to begin the next stride (Pearsall et al., 2000).

Common to starting accelerations in forward skating is the transition from running-like strides to gliding strides (de Koning et al., 1995). Each push-off produces explosive power to help accelerate an individual's center of mass (CoM) with respect to the ground (Van Ingen Schenau, de Boer, & Groot, 1989). The running-like strides are essential for accelerating up to and reaching top speeds. The best time-based linear performances have typically been found associated with a shorter glide phase to maximize time spent in the force producing phase of the skating stride and in turn, maximize stride rate (Culhane, 2012; de Koning, de Boer, de Groot, & Schenau, 1987). In a straight-line acceleration this powerful sprint technique lasts only a few strides before full strides take over (Lafontaine, 2007; Pearsall et al., 2013; Renaud et al., 2017).

1.2.3 Motion Capture in Skating

Investigating the movement intricacies of ice hockey is a unique challenge in sport biomechanics research. Early kinematic studies have been conducted in the sport of speed skating where forward movement paths are predictable about the circuit allowing for defined fields of view for motion capture. Contrarily, the study of ice hockey kinematics advanced at a slower rate due to the complexity of capturing accurate skill simulations as result of extensive motion variations, large movement areas, and the difficulties of setting up a motion capture system on an ice rink (Lafontaine, 2007; Upjohn et al., 2008). Early ice hockey studies provided insight into skating using two-dimensional (2D) video analysis as means of quantifying body kinematics in forward starts and full strides (Marino, 1977; Marino, 1979).

Techniques for obtaining three-dimensional (3D) kinematic analysis of body movement in hockey have been more challenging. With recent advancements in optical tracking and body worn sensor technologies, more detailed kinematic investigations were able to be conducted over the past two decades. For example, small gage electrogoniometers paired with a light, portable data logger has been successfully used in several ice hockey skating studies to allow free movement of skating subjects over large areas of the ice surface with direct and concurrent tracking of hip, knee and ankle joints (R. Chang, 2003; Pearsall et al., 2001; Stidwill, Pearsall, & Turcotte, 2010). Novel multi-tracking camera cinematography approaches to capture three-dimensional motion in hockey over an ice surface have be explored. For instance, Lafontaine (2007) demonstrated a method of on ice motion capture using two high speed cameras and a guide rail system to obtain lower limb kinematics during forward acceleration tasks; allowing for a multi-stride analysis of lower body joints. Other approaches have used a stationary skating treadmill and multiple cameras to obtain 2D or 3D body kinematic data of forward skating (Lockwood & Frost, 2007; Upjohn et al., 2008).

More recently, state-of-the-art camera systems have allowed for on-ice motion capture in the rink environment bringing ice skating research one step closer towards analyzing unobstructed, natural movement. This feat has been accomplished using optical tracking of passive reflective markers on a skater to track their body kinematics. These latest ice skating kinematic studies have been able to accomplish in depth analyses of numerous skate strides on ice, providing detailed lower body and full body segment motion descriptions (Hallihan, 2018; McPhee, 2018; Renaud et al., 2017; Shell et al., 2017; Van Der Kruk, Schwab, Van Der Helm, & Veeger, 2016). Currently, comparisons have been drawn between high and low calibre ice hockey players (Renaud et al., 2017) along with male-female comparisons for forward skating (Shell et al., 2017) and stop-and-go tasks (Hallihan, 2018). These on-ice 3D biomechanics studies have demonstrated the feasibility of capturing precise data from distal blade-to-ice angles in a global reference frame to local and global full body joint angles. It has also been shown that kinematics can be captured simultaneously with other measurements in a natural playing environment as seen with employment of innovative player equipment (McPhee, 2018) or electromyography (Buckeridge et al., 2015; Hallihan, 2018) alongside optical tracking systems.

1.2.4 Electromyography in Sport

Quantification of the physiological process in which muscles produce force has become feasible through the employment of electromyography (EMG) systems (De Luca, 1997). Surface electrodes in particular have provided a non-invasive means of measuring this feedback (De Luca, 1997). This technology delivers valuable insight into individual muscle activity amplitudes and temporal information that have application in the biomechanics research field. These details can in turn potentially be used to comprehend muscle synergies or compare muscle coactivation during a task.

Analysis of muscle activity has been frequently examined in the running biomechanics field. For instance, EMG has been used to observe the impact of distance and pace on muscle activity during running. Findings have indicated that the muscles responsible for movement at the hip joint became fatigued faster than other lower body muscles (Hanon, Thépaut-Mathieu, & Vandewalle, 2005). An increase in lower body muscle EMG amplitude was also identified with respect to an increase in running speed (Kyröläinen, Avela, & Komi, 2005). Similarly, runner fatigue has been investigated in relation to lower body muscle activity and its effect on joint stiffness and the stretch-shortening cycle (SSC) in response to speed intensities (Komi, 2000). Of note was the change in leg stiffness regulation with exhaustive SSC fatigue where a reduction in stretch-reflex sensitivity and muscle stiffness were found to have a negative impact on the force producing mechanisms. Ignoring fatigue effects, the body has been discovered to adapt the spring-like behavior of the legs at faster running velocities through a combination of maintaining and increasing joint stiffness reflected in EMG data (Kuitunen, Komi, & Kyröläinen, 2002).

Employment of EMG in other sports has been used to assess high intensity sprint simulations on anaerobic and aerobic power measures. It has been indicated that muscle activity amplitude decreases during short succession sprint repetitions (Mendez-Villanueva, Hamer, & Bishop, 2008) or single sustained bouts, while adequate rest allowed for full muscle recovery (Vesterinen, Mikkola, Nummela, Hynynen, & Häkkinen, 2009). When it comes to investigations of ice skating sports, EMG has predominantly focused on describing muscle activation patterns in regard to optimal movement conditions. Within the field of speed skating, a number of earlier studies defined the lower limb muscle activity of high calibre athletes (De Boer et al., 1987; de Koning et al., 1991) through capture of superficial muscles spanning the hip, knee, and ankle joint. De Boer et al (1987) indicated that push-off power spawned predominantly from the extensor muscles of the Gluteus Maximus and Vastus Medialis with the later study (de Koning et al., 1991) supporting this conclusion. De Koning et al (1991) further added the findings of proximal to distal temporal muscle recruitment in the lower limbs during push-off. With respect to ice hockey biomechanics, the earliest studies utilizing EMG systems examined the effect of skating speed on lower body muscle activity during forward motion on a skating treadmill. Findings reflected those of aforementioned running speed studies where an increase in skating speed caused an increase in muscle EMG amplitudes (R. Chang, Turcotte, & Pearsall, 2009; Goudreault, 2002). Additionally, identification of the Adductor Magnus muscle's integral role in hip abduction-adduction for power production in response to increased skating speeds was also highlighted during the later treadmill study (R. Chang et al., 2009). Buckeridge et al (2015) captured muscle activity on ice during forward accelerations as a component of their study to draw comparisons between acceleration and steady state strides. Through direct comparison of all 5 muscles for each stride type, the Medial Gastrocnemius was found to be more activated during the acceleration stride versus the steady state stride while the Vastus Lateralis and Medialis presented the reverse. These findings can be explained in relation to knee range of motion (ROM) changes during acceleration as ROM increased with each stride due to greater knee extension occurring at steady state compared to the acceleration phase.

1.2.5 Exoskeletons

The emergence of performance assistive technologies to work in synchrony with the human body has become increasingly prevalent in various industries and research fields. Commonly termed exoskeletons, these technologies can be identified into one of two types according to their properties; active exoskeletons or passive exoskeletons (van Dijk, 2015). As suggested by the names, active exoskeletons draw upon an external source of power working alongside the human system to input additional energy while their passive counterparts rely upon harmonious integration with the human system to optimize motion using exclusively human generated power. Each type has unique benefits, however, passive designs offer more simplistic means of obtaining motion augmentation without the necessity for overly burdening equipment. Within this grouping of passive exoskeletons there are two sub classifications based upon the employment of rigid elements such as springs or soft elements such as elastics.

Numerous passive exoskeletons have been conceptualized and fabricated for the applications of hopping, walking, and running with varied results. Emphasis of these investigations are commonly placed on measures of joint stiffness, metabolic power or metabolic consumption as indications of successful system-user interaction and motion optimization. Though not directly considered an exoskeleton, an investigation of compression clothing on vertical jump height performance found that apparel design to alter compression properties or hip stiffness via elastic materials benefitted user jump results (Wannop et al., 2016). This study suggested that the performance augmentation induced by clothing passive elements was likely related to alterations in muscle force-length or force-velocity relationships. Operating on the concept of joint stiffness modification, use of novel spring-leaf exoskeletons working in parallel with a user's full lower body has been found to decrease net metabolic power (Grabowski & Herr, 2009). These results suggest that the exoskeleton system functioned through the effective storage and return of potential energy generated by the user's weight, which in turn allowed for a reduction in metabolic cost. However, the authors indicated the need for an investigation of muscle activity while wearing the suit to better comprehend the mechanisms behind the reduction in metabolic cost. Examination of the soleus muscle was conducted for a different hopping study in which participants wore a passive spring exoskeleton working in parallel with the ankle muscle tendon unit (Farris et al., 2013). The additional element reduced Soleus muscle activity as anticipated along with muscle force and net metabolic power. Interestingly, mechanical work remained unchanged by the system.

Exoskeletons designed for walking and running locomotion aid have been predominantly conceptualized for the reduction of metabolic cost in lieu of power production. Simulations of walking locomotion using theorized singular and multi-joint spanning exotendons have indicated substantial reductions in joint moments and power, suggesting a reduction in metabolic costs (van den Bogert, 2003). When experimentally investigated, the tri-articular design did yield a reduction in average absolute joint torques, yet observed alterations in joint kinematics undermined the simulation's conclusion surrounding metabolic benefits (Van Dijk & Van Der Kooij, 2014). More recently, implementation of a passive ankle exoskeleton containing a clutch to disengage the spring element was able to achieve a reduction in metabolic walking cost at an optimal system stiffness (Collins et al., 2015). The discovered reduction in calf muscle force while maintaining normal ankle function as demonstrated by Soleus EMG feedback, formulated an association between lowered muscle activity and walking metabolic efficiency. Jackson and Collins (2015) used a powered ankle exoskeleton to assist walking through a chosen input of work or torque as means of assessing their independent effects on a user. Though both showed a reduction in effort-related measures, only work input reduced metabolic cost where torque input lead to an increase (Jackson & Collins, 2015). Since then, a number of other studies have also designed and demonstrated reductions in energy via indirect calorimetry while walking with autonomous, low profile powered exoskeletons (Mooney & Herr, 2016; Panizzolo et al., 2016; Seo, Lee, & Park, 2017).

Running with assistive systems has been investigated to a much lesser extent. A passive knee exoskeleton utilizing carbon fibre spring elements and a clutch mechanism identified the need for careful planning of exoskeleton design to avoid any system obstructions on natural human segment movement (Elliott et al., 2013). This study also noted the discomfort of bulky and heavy equipment during activity, highlighting the necessity for consideration of user-equipment interactions to avoid counterproductive performance alterations. Lee et al. (2017) used an offboard actuator to control assistance profiles of a power hip exosuit during running, demonstrating that a metabolic cost reduction can be obtained through worn sporting apparel. Of interest in this study was that the simulation-optimized profile outperformed the biological hip driven profile through suspected indirect changes of other lower limb joint movement. This finding remains to be verified by muscle activity.

To our knowledge only two studies have strictly detailed the response of lower limb muscle activity waveforms to an exoskeleton during human locomotion. Steele et al. (2017) looked to enhance future exoskeleton design by investigating ankle exoskeleton impact on muscle recruitment and coordination that may have played a role in the alteration of metabolic cost and kinematics during walking. Reductions in the Soleus and Gastrocnemius muscles were found with increasing exoskeleton aid, yet co-contraction patterns notably remained similar whether walking with or without the exoskeleton. Furthermore, it was found that the same three muscle synergies explaining 94.5% of the variability in the control condition could explain significantly less variability when the exoskeleton was employed. Muscle coordination complexity was similar between assisted and unassisted walking, however, either pattern structure or activation were being affected by the system, especially for the ankle plantar flexor driven synergy (Steele et al., 2017). Correlations between individual muscle activity and changes in metabolic rate were also found, with quadricep activity being the leading predictor for both walking conditions (Steele et al., 2017). Ferris et al. (2005) investigated planter flexor muscle activity in relation to a hopping task so to better interpret human adaptation of joint stiffness in reflection of an added spring element. Overall leg stiffness was found to be unaltered by the worn exo-system yet biological ankle stiffness was found to be lower in the spring condition. Muscle activation amplitudes were found to be lower for six of eight recorded leg muscles during stance phase when the spring was attached to the orthosis compared to without (Ferris et al., 2005). These findings supported the concept that local stiffness of the system targeted joint was reduced to accommodate the parallel element and in turn overall leg stiffness. The adjustment of biological ankle stiffness was suggested to stem from the found reduction in plantar flexor muscle activity (Ferris et al., 2005).

1.3 Objectives & Hypothesis

The objective of the current study was to assess the effects of a passive hip exotendon suit on lower limb muscle activity of participants during forward skating accelerations. The goal of this study was to 1) identify if and to what extent the exotendon suit can affect lower limb muscle function and 2) to better understand the changes in muscle recruitment that occur during the skate start by comparing stride-to-stride surface electromyography (EMG, mV) measures of lower limb muscle recruitment patterns within control and tension band conditions.

EMG measures were evaluated by 1) comparing within muscle Root Mean Square (RMS) activity for each stride interval during the entire Skate Start trial and by 2) Principal Component Analyses (PCA) of hip extensor muscle's EMG pattern for one early and late Skate Start stride.

The specific hypotheses were:

- The passive exotendon system will cause changes in muscle activity between conditions. Specifically:
    - Hip extensor muscle activity will decrease with increasing band stiffness during push-offs throughout the acceleration task
- Gross differences will be found in muscle activity over a skating start from early 'Acceleration' strides to later 'Steady State' strides. Specifically:
    - Hip abductors and knee extensors muscles will be more activated during steady state strides
    - Hip extensors will remain consistent
    - Knee flexor and plantar flexor muscles will be more activated during acceleration strides

Chapter 2: Methodology

2.1 Participants

Nine male hockey players were recruited for this study. All participants were considered high caliber with a minimum competition level of the Canadian U Sports league or above. The participants were between the ages of 21 to 33 years old and were free of any lower body injuries during the collection period of the study (Table 1). The surface EMG data collected was obtained as part of larger study approved by the McGill University Research Ethics Board II (REB File #: 82-0717). Sample size was dictated by the preceding research project which indicated that earlier hockey studies were able to yield sufficient statistical power with similar numbers (McPhee, 2018).

Table 1. Original participant demographics

	Mean (± SD)
Age (years)	24.11 (3.54)
Height (m)	1.80 (0.07)
Weight (kg)	85.40 (7.97)
Playing experience (years)	19.80 (3.42)

2.2 Instrumentation

Anthropometric participant data were recorded using a tape measure and force plate (Type 9260AA, Kistler AG, Winterthur, Switzerland). Each participant had their dominant leg (i.e. preferred kicking leg side) outfitted with 6 wireless surface EMG sensors (DTS EMG

Sensor part #542, Noraxon U.S.A. Inc., Scottsdale, AZ) connected to disposable, self-adhesive Ag/AgCL electrodes (Noraxon Dual Electrodes Product #272S, Noraxon U.S.A. Inc., Scottsdale, AZ) in accordance to SENIAM guidelines (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000). The disposable electrodes had an inter-electrode spacing of 2.0cm. Electrodes were placed over the following muscles: gluteus maximus (GMA), gluteus medius (GME), adductor group (ADD), biceps femoris (BF), vastus medialis (VME), and medial gastrocnemius (GAS). Preceding electrode placement, electrode sites' hair was removed with a disposable razor. No application of conductive gel or cream was necessary for the EMG system (Noraxon TeleMyo2400T G2DTS System, Noraxon U.S.A. Inc., Scottsdale, AZ). Placement sites were located on the midline of the muscle bellies, half way between the myotendinous junction and muscle belly origin. Electrodes were placed parallel to the estimated muscle fibre directions (De Luca, 1997). Plastic tape was used to help secure electrodes. Compression pants were carefully pulled on over top of the lower body EMG setup followed by a tight fitting full-body Velcro motion capture suit (NaturalPoint, Inc. DBA OptiTrack, Corvallis, OR) to further minimize electrode movement.

Surface EMG signals were wirelessly recorded using a TeleMyo 2400T G2 DTS Belt Receiver (Noraxon U.S.A. Inc., Scottsdale, AZ) sampling at 1500Hz with a gain of 500Hz. Recordings were synced with body kinematic data by way of Nexus software (Vicon, Oxford, UK) as an analog input devices. The sensor delay of 36ms in the EMG system was adjusted in Nexus using the 'Delay Compensation' tool.

The exotendon suit fabrication details and design choices have been outlined in the simultaneous motion capture study completed by McPhee (2018) (Appendix A). To briefly summarize, the exotendon system was created to target the largest power generating joint during skating accelerations: the hip. Powered mechanisms were avoided to 1) reduce system burden on participants and 2) hinder integration potential with current or future hockey equipment. Instead to assist hip joint extension, a light weight climbing harness (Momentum Harness, Black Diamond, U.S.A.) was modified by adding aluminum cam buckles to securely hold latex tension bands posterior to a subject's hip joint thereby acting as an extra elastic unit. This "exotendon" was adjustable to subject's size and permitted direct, rapid interchange of tension bands between testing trials. Three pairs of bands (Power Guidance Fitness Training, Suzhou, China) were chosen based on their physical properties to create three band stiffness conditions (Appendix A&B. The estimated change in length of the bands during hip flexion between 0.11 and 0.23m. Soft (red band) = 132.2 N/m; Medium (black band) = 256.7 N/m; Stiff (purple band) = 440.8 N/m). Participants wore the system atop of Velcro suit and surface EMG setup with careful planning of harness strap locations and band paths to ensure no tampering was occurring with the electrodes, leads, or wireless EMG transmitter units.

Figure 1. The posterior view of the exotendon system employing the stiffest elastic condition (STAP).
*Original figure created by and presented with the permission of Brian McPhee (McPhee, 2018)

Participants were supplied testing skates (Bauer Vapor 1X 2016 Model, Bauer Hockey Ltd., Exeter, NH, USA) to wear during testing. Subjects were asked to bring their own skate blades sharped to their personalized preference. A hockey stick was also provided to the subject to skate with to best mimic an in-game situation. Participants were instructed to lace the skates as they would normally for competition.

2.3 Experimental Protocol

2.3.1 Off-ice subject testing preparation

Upon arrival to the ice arena change room, participants were given a verbal overview as well as written instructions of the testing protocol. Participants were asked to sign the study's consent form. A personal information form was then provided for participants to complete which included injury histories, player experience, and limb dominance. Time was allocated to answer protocol related questions asked by the participant to the researchers. The participants were subsequently brought into an adjacent private room in which the EMG sensors were placed on the dominant leg and all testing apparel put on. Surface EMG placements were only completed on the participant's dominant leg defined as the leg with which the participant would choose to kick a ball (van Melick, Meddeler, Hoogeboom, Nijhuis-van der Sanden, & van Cingel, 2017). Once dressed in lower body compression pants (to minimize EMG sensor shifting), the participant was brought back into the change room where the exotendon system was pulled on. At this time, band length was normalized for each participant to 35% of leg length with tape markings applied to the bands for quick reference. Skate blades were exchanged with the participant's chosen testing skate size pair and given to the participant to lace up.

2.3.2 On-ice Participant Testing Procedure

Each participant was brought to the ice surface where they were given 5 minutes to warm up and familiarize themselves with the testing skates and harness system without any band. After warmup, a check of the surface EMG system's functionality was conducted by having participants preform a one-legged squat on their dominant leg. The first bands were then secured on by the cam buckles in accordance to the participants condition order. The bands were pulled through the cam buckles at each end up to the tape markings for proper individualized length adjustment. Band order was randomized for each participant including a control condition where the harness was worn without any bands in place. Once the bands were secured, participants performed skating start acceleration tasks over 15 meters from a forward-facing ready position. The start line and finish lines were clearly identified by two sets of orange cones. Participants were free to choose their start skating's lead leg side as they preferred. Three trials were consecutively completed for each condition with one-minute rest between trials for participant recovery as well as inspection of the band length. After three accelerations, participants were allotted a 5-minute rest period during which band conditions were substituted.

2.4 Data Analysis

All the data collected with the VICON Nexus 2.2.5 software was inspected for quality of the EMG data to verify that participant data was usable. Two participants were removed from the study due to problematic sensor feedback resulting in non-existent or unfilterable signals. After removing the problematic data, the remaining 7 participants data sets were run trial-by-trial through a custom Visual3D (Ver 5.01.23, C-Motion, Germantown, Maryland, United States) pipeline. EMG signal data was filtered using a bandpass filter created by applying a 6^{th}-order zero-lag high-pass Butterworth filter at a cut-off frequency of 20Hz and a 6^{th}-order zero-lag low-pass Butterworth filter at a 500Hz cut-off frequency. The signal waveform was then full wave rectified and smoothed by a 200-point moving average window. Identification of locomotion "ON" and "OFF" events (Hreljac & Marshall, 2000) were accomplished by way of kinematic proxy measures in this software i.e. Skate "ON" and "OFF" the ice were defined as ON = toe marker maximal vertical acceleration, and OFF = the jerk of the toe marker across zero in the anterior-posterior direction (Hreljac & Marshall, 2000). These skate ON and OFF events were detected for each skate stride cycle within a trial. All trials underwent manual inspection within the Visual3D processing pipeline to confirm correct event identification. Upon completion of trial processing in Visual3D, the data were imported into MatLab (version r2018b, Mathworks, Natick, Massachusetts, U.S.A.) for further analysis.

Custom MatLab scripts adapted from the biomechZoo toolbox (Dixon, Loh, Michaud-Paquette, & Pearsall, 2017) were used to complete the data analysis. Initially, left and right limbs were identified as “Side1” and “Side2” where “Side1” was the leg on which the initial step occurred. Each step (S) within “Side1” and “Side2” were designated by odd and even step numbers, respectively. Skate “ON” and “OFF” events (define in Visual3D) were subsequently tagged with each step (Figure 2). EMG data was then down sampled from 1680 Hz to 240 Hz so as to match the concurrent studies’ motion capture system (tracking body movement). At this moment, a trial event check was executed to determine the furthest step event commonly available by participants. A majority of the trials across all participants were found to share events up to S7ON (with exception of 3 trials from exclusive participants and conditions). To maintain maximal available events in the studies data set, these 3 trials were removed from the analysis. A check was also done to verify whether the participant used their dominant or non-dominant leg as “Side1”. Since EMG electrodes were placed exclusively on the dominant leg, two of the seven remaining participants were removed from further analysis as their “Side1” limb was on their declared non-dominant side.

For each of the remaining participants, muscle EMG waveform magnitudes were then normalized to a Maximum Dynamic Activation (MDA) value for each respective muscle across all conditions within a participant. This normalization process was chosen in reflection of the work by Albertus-Kajee et al. (2011) which suggested a MDA EMG normalization approach for investigating maximal muscle activity was optimal during running investigations, rather than Maximum Voluntary Contractions (MVC) because RMS magnitudes in dynamic movements often supersede MVC thresholds. As such MDA values for trial normalization were derived for each participant's muscles exclusively.

Lastly, Stance phases (ST) were then created between skate ON and skate OFF events for each leg side: i.e. S1ON to S3OFF was defined as ST1 (i.e. stance 1), S2ON to S4OFF as ST2, and so on up to ST6 (Figure 2). The root mean square (RMS) activity of each muscle's EMG waveform for each stance event was extracted.

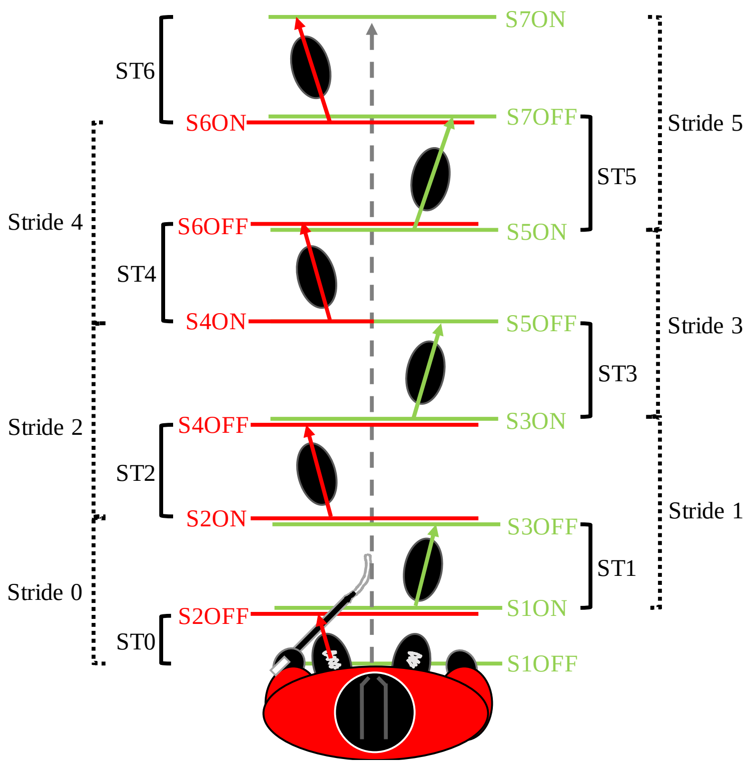

Figure 2. Skating events during a forward acceleration task (as seen from top view of a Skating Start's footstep sequence and corresponding stance phases). "ON" and "OFF" events denote the skate contacting and leaving the ice. Green events are for the Side1 limb while the red events are for the Side2 limb. "ST" events are stance phases where the skate is on the ice through the glide and push-off. Stride events show the entire stride from initial ice contact by the skate through both stance (blade on ice) and repositioning (blade off the ice) phases to the next time the skate contacts the ice.

2.5 Principal Component Analysis

Principal Component Analysis (PCA) permitted for the identification of important waveform characteristics while simultaneously reducing the dimensionality of EMG data (Deluzio & Astephen, 2007). Four PCA analyses were conducted in total. Each analysis respectively investigated the stance phase of an acceleration (ST1) and steady state stride (ST5) for the two muscles that directly affect hip flexion/extension: the Gluteus Maximus and the Biceps Femoris. The decision to run these analyses in separate PCAs for each muscle for one selected stride was to allow for variability between conditions to be addressed in line with the studies objectives. The muscles included were specifically chosen based on their direct functional relationship with the exotendon system related to hip extension. This analytical approach further helps in the identification of critical moments within a skating push off where muscle activity may be subject to large or small variability. To compute each analysis, data was entered into a matrix (**X**) of *n* x *m* where *n* represented the number of trials for every participant and *m* was the 101 points of data collected over the stance phase of the stride cycle. Trials were input individually into each analysis instead of input by way of ensembled averages as means of increasing matrix size for a more stable analysis. All matrices (**X**) were 58x101. From the covariance matrix **X** that was created, eigenvectors, also known as principal components (*PC*), were extracted. This element represents important waveform characteristics including shape, magnitude, timing shifts, or difference operators. Eigenvalues computed for each *PC* indicated the amount of variability present in the *PC*. The total number of *PCs* examined for these analyses were chosen based upon a threshold of 90% which ensures the majority of total data set variance was captured (Chau, 2001; McKean et al., 2007). This meant that all *PCs* cumulatively required to explain 90% of the dataset’s variability were kept. For all four analyses this came to 3 *PCs*.

Principal component scores (*PC-scores*), representing a waveforms likeness to the *PC* shape, were derived last (*PC-scores* = $(\mathbf{X}-\bar{\boldsymbol{X}})*PC$). These *PC-scores* were used in proceeding statistical testing as primary dependent variables. PCA analyses were performed in MatLab.

2.6 Statistical Analysis

Statistical tests were performed in SPSS (Version 23.0, IBM Corporations, North Castle, NY, USA). RMS EMG activity was compared for each stride phase common through all acceleration trials by way of band condition groups (STAC, STAR, STAB, STAP). *PC-scores* were similarly compared between condition groups for the individually chosen phases. Trial RMS values and *PC-scores* were averaged within participant for each band condition before being entered into any statistical testing.

Due to small sample size in the study, interpretation of data normality was conducted to decide on parametric or non-parametric testing. Shapiro-Wilks tests of normality were run on data once grouped into respective band conditions. Data distribution skew and kurtosis was also verified to formulate a decision regarding normality. The vast majority of band condition data presented as normally distributed; hence, parametric tests were chosen for continuation of the analysis. Mixed-ANOVAs were run with Bonferroni corrections to test for interactions between RMS EMG activity of each band condition (STAC, STAR, STAB, STAP) and stance phase (ST1 to ST6) for each muscle (ADD, GMA, GME, BF, VME, GAS).

To compared band condition *PC-scores*, one-way repeated measures ANOVAs were used as dictated by the PCA analyses which were performed for individual stance phases. *PC-scores* were also plotted on principal component by principal component (*PC x PC*) plots to visually inspect for score clustering. Sphericity was tested for via Mauchly’s Test and a Greenhouse-Geisser correction was performed to adjust the degrees of freedom (df) if the assumption of sphericity was violated. Levene's test for equality of variances was simultaneously run to ensure the assumption of equal variances held true. An alpha level of 0.05 was set for all tests.

Chapter 3: Results

3.1 Demographics and Elastic Properties

During data processing, four of the participants from the original collection were removed. The remaining participant sample demographics were re-examined as found in Table 2.

Table 2. Participant demographics (*N*=5)

	Mean (± SD)
Age (years)	26.4 (4.72)
Height (m)	1.79 (0.05)
Weight (kg)	83.0 (6.31)
Dominant leg length (m)	0.93 (0.03)
Playing experience (years)	20.4 (4.83)

Band material properties were examined in the concurrent study by McPhee (2018) (Appendix B). Band stiffness properties quantify the red band (STAR) as the least resistive or soft condition, the black band (STAB) as the mid resistive or medium condition, and the purple band (STAP) as the most resistive or stiff condition. Time performance to accelerate 9 meters under each condition was also deduced in the 2018 study including an evaluation of best and worst conditions (Appendix C), showing that the system neither benefitted nor hindered participant.

3.2 Muscle Activity and the Exotendon System

When assessing lower body muscle activity of the six muscles for each band condition over the course of an acceleration, no significant interactions were found between elastic condition and stance phase for RMS muscle activity (Table 3). Furthermore, no significant main effect of elastic condition was produced when elapsing RMS EMG activity across the whole trial (Table 4).

Table 3. Root mean square (RMS) electromyographic (EMG) activity and standard deviation (±SD) of 6 lower body muscles for 3 band (STAR, STAB, STAP) and control (STAC) conditions during 6 consecutive stance phases of a forward skating acceleration task. Numbers are expressed as a percentage of muscle Maximal Dynamic Activation.

Muscle	Condition	ST0 RMS activity (±SD)		ST1 RMS activity (±SD)		ST2 RMS activity (±SD)		ST3 RMS activity (±SD)		ST4 RMS activity (±SD)		ST5 RMS activity (±SD)		ST6 RMS activity (±SD)	
Adductor Group	STAB	8.4	(4.5)	17.1	(10.1)	29.7	(16.5)	25.1	(16.7)	33.0	(17.5)	23.8	(12.4)	28.0	(16.3)
	STAC	6.9	(4.8)	16.3	(11.1)	30.3	(11.6)	23.9	(13.3)	31.3	(15.9)	23.0	(13.8)	26.2	(15.3)
	STAP	7.6	(4.0)	18.8	(15.8)	28.9	(11.7)	24.0	(13.3)	29.2	(16.6)	21.5	(11.9)	27.5	(15.0)
	STAR	7.5	(7.8)	15.2	(10.1)	28.5	(18.1)	23.3	(11.9)	33.7	(13.5)	19.9	(10.6)	27.2	(15.3)
Gastrocnemius	STAB	10.5	(12.9)	18.5	(5.2)	35.5	(9.5)	22.1	(19.6)	37.1	(7.3)	16.1	(9.6)	34.7	(5.9)
	STAC	10.4	(10.0)	16.9	(6.0)	31.3	(13.1)	14.4	(9.0)	40.5	(9.3)	13.9	(11.1)	38.1	(6.0)
	STAP	11.2	(11.1)	18.2	(5.2)	33.8	(8.5)	15.0	(8.3)	35.9	(7.8)	13.4	(7.3)	34.0	(8.8)
	STAR	11.5	(13.7)	17.6	(8.4)	37.3	(10.9)	20.3	(15.0)	36.2	(6.8)	19.3	(17.7)	33.2	(5.3)
Gluteus Maximus	STAB	7.4	(8.5)	30.8	(15.4)	19.1	(12.1)	25.7	(12.8)	19.6	(12.5)	30.1	(13.1)	20.6	(11.9)
	STAC	9.7	(12.0)	33.9	(17.1)	22.4	(12.2)	32.4	(15.7)	21.6	(13.4)	32.6	(19.0)	26.7	(12.0)
	STAP	7.6	(8.2)	30.7	(16.7)	24.1	(9.6)	29.1	(15.0)	20.5	(12.4)	30.9	(7.8)	19.9	(10.3)
	STAR	6.9	(8.6)	32.6	(16.4)	17.8	(11.1)	32.7	(18.2)	21.6	(13.7)	32.6	(15.5)	23.9	(15.1)
Gluteus Medius	STAB	23.1	(9.8)	32.9	(9.1)	23.1	(11.5)	39.0	(7.9)	30.6	(7.8)	40.2	(3.8)	35.4	(8.9)
	STAC	25.7	(9.1)	33.7	(8.4)	26.1	(9.6)	38.4	(9.7)	30.5	(8.6)	40.5	(5.5)	35.5	(9.8)
	STAP	22.0	(5.2)	30.7	(5.6)	24.0	(10.3)	41.3	(7.6)	30.7	(11.1)	41.1	(3.4)	32.4	(8.3)
	STAR	24.7	(8.7)	33.8	(5.8)	28.6	(14.8)	39.5	(3.8)	32.6	(6.0)	41.7	(6.9)	32.7	(9.2)
Biceps Femoris	STAB	11.7	(9.1)	33.6	(13.8)	31.6	(13.9)	35.5	(6.3)	29.8	(14.0)	27.6	(9.0)	31.3	(18.5)
	STAC	12.0	(7.7)	29.7	(18.6)	26.2	(9.7)	34.3	(13.1)	30.6	(13.7)	30.0	(14.8)	29.8	(15.2)
	STAP	12.2	(9.5)	27.3	(15.5)	25.5	(11.5)	34.5	(14.5)	29.6	(13.4)	26.9	(13.0)	26.0	(15.1)
	STAR	12.7	(8.3)	35.2	(17.3)	31.9	(12.5)	32.9	(7.5)	30.5	(13.8)	21.1	(9.3)	29.9	(18.4)
Vastus Medialis	STAB	11.1	(7.9)	32.9	(15.1)	17.7	(12.7)	32.9	(17.7)	26.0	(13.0)	31.9	(17.8)	26.0	(13.5)
	STAC	7.9	(4.7)	40.5	(7.6)	20.2	(13.1)	38.9	(10.6)	26.9	(10.5)	36.3	(10.4)	29.9	(9.6)
	STAP	8.9	(6.0)	41.5	(9.8)	19.5	(11.9)	39.7	(11.4)	26.5	(11.2)	34.4	(11.3)	25.5	(6.2)
	STAR	7.2	(6.9)	35.2	(14.5)	16.5	(14.6)	33.6	(15.0)	22.1	(9.6)	31.4	(15.8)	21.3	(10.3)

Table 4. Root Mean Square (RMS) electromyographic (EMG) activity and standard deviation (±SD) of 6 lower body muscles during a forward skating acceleration task under 3 band (STAR, STAB, STAP) and control (STAC) conditions.

Muscle	Condition	Entire trial RMS activity (±SD)	
Adductor Group	STAB	18.5	(8.4)
	STAC	17.4	(8.3)
	STAP	17.5	(7.9)
	STAR	17.3	(8.2)
Gastrocnemius	STAB	17.5	(11.2)
	STAC	16.4	(11.5)
	STAP	15.6	(10.8)
	STAR	18.1	(10.5)
Gluteus Maximus	STAB	17.1	(7.5)
	STAC	20.1	(8.4)
	STAP	17.3	(8.6)
	STAR	19.0	(8.6)
Gluteus Medius	STAB	20.2	(13.2)
	STAC	20.8	(13.2)
	STAP	19.5	(13.8)
	STAR	20.6	(14.0)
Biceps Femoris	STAB	20.4	(10.6)
	STAC	20.4	(9.2)
	STAP	19.6	(8.2)
	STAR	20.1	(10.1)
Vastus Medialis	STAB	19.7	(8.6)
	STAC	19.1	(12.8)
	STAP	18.9	(12.4)
	STAR	18.2	(9.4)

Results did reveal a main effect of stance phase on muscle activity when ignoring the condition factor during skating accelerations (Table 5). Odd number stance phases captured dominant leg push-off activity while even number phases reflected dominant leg activity during the non-dominant leg push-offs. Results will be focused primarily on the odd number stance phases (ST1, ST3, ST5) during which stride power production was occurring (Figure 3a, 3b). The Adductor group saw a significant increase in RMS muscle activity from ST1 to ST3 by 7.24

%MDA ($p=0.002$) and by 5.19 %MDA to ST5 ($p=0.038$). A similar trend was found for the gluteus medius RMS activity with an increase of 6.77 %MDA from ST1 to ST3 ($p=0.009$) and 8.09 %MDA from ST1 to ST5 ($p=0.001$). Neither of these muscles showed significant differences between ST3 and ST5 phases. The Biceps Femoris activity displayed a unique pattern with a significant difference appearing between ST3 and ST5 as RMS activity dropped by 7.93% ($p = 0.003$). As well, significant differences between stance phases of vastus medialis was found with a drop of 4.02% from ST1 to ST5 ($p=0.024$). The gluteus maximus and the gastrocnemius muscles were found to have remained similarly activated throughout the three push-off stance phases captured ($p>0.05$ for ST1, ST3, ST5).

Table 5. Root Mean Square (RMS) electromyographic (EMG) activity and standard deviation (±SD) of 6 lower body muscles during 6 stance phases during a forward skating acceleration task. RMS activity values have been collapsed across the condition factor. Green columns indicate push-off phases for the dominant limb. White columns reflect the dominant limb activity during the time which the non-dominant limb is in stance phase.

Muscle	ST0 RMS activity (±SD)	ST1 RMS activity (±SD)	ST2 RMS activity (±SD)	ST3 RMS activity (±SD)	ST4 RMS activity (±SD)	ST5 RMS activity (±SD)	ST6 RMS activity (±SD)
Adductor Group	7.6 (1.7)	16.8 (2.7)**	29.4 (3.3)	24.1 (2.1)*	31.8 (1.7)	22.0 (1.3)*	27.3 (0.6)
Gastrocnemius	10.9 (1.6)	17.8 (1.5)	34.5 (2.0)	17.9 (5.3)	37.4 (1.1)	15.7 (4.5)	35.0 (1.6)
Gluteus Maximus	7.9 (1.8)	32.0 (0.7)	20.8 (1.2)	30.0 (2.2)	20.8 (0.7)	31.5 (4.7)	22.8 (2.0)
Gluteus Medius	23.9 (2.0)	32.8 (1.7)**	25.4 (2.3)	39.5 (2.5)*	31.1 (2.1)	40.9 (1.6)*	34.0 (0.6)
Biceps Femoris	12.1 (0.8)	31.5 (2.1)	28.8 (1.8)	34.3 (4.0)*	30.2 (0.3)	26.4 (2.8)*	29.3 (1.9)
Vastus Medialis	8.8 (1.4)	37.5 (3.7)*	18.5 (1.1)	36.3 (3.3)	25.4 (1.4)	33.5 (3.5)*	25.7 (3.0)

* Indicates a significant difference with one other stance phase executing a push-off ($p<0.05$)

** Indicates a significant difference with two other stance phases executing a push-off ($p<0.05$)

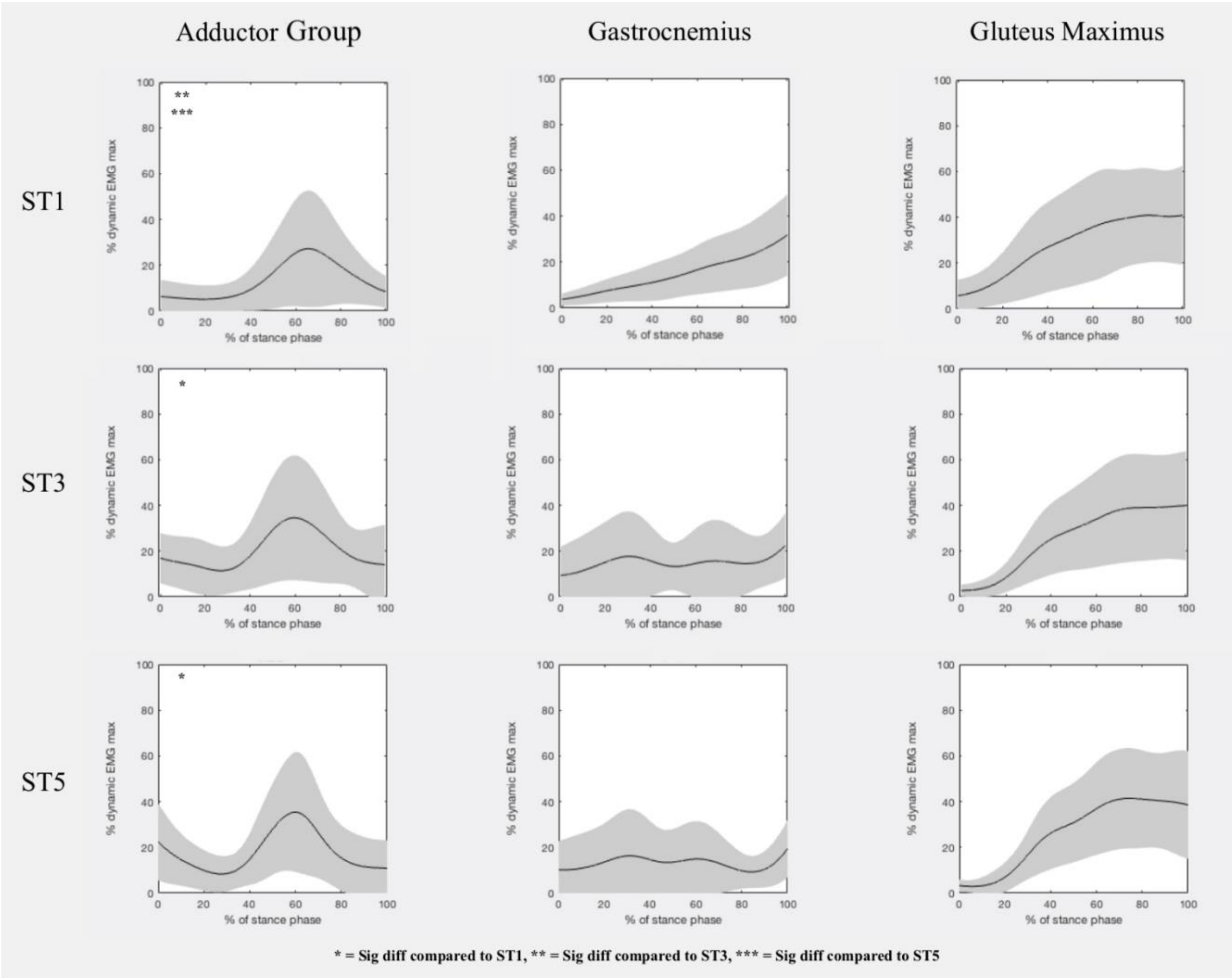

Figure 3a. Push-off stance phase Root Mean Square (RMS) electromyographic (EMG) activity waveforms (ST1, ST3, ST5) for Adductor Group, Gastrocnemius, Gluteus Maximus muscles during a forward skating acceleration task.

The black line indicates the mean of participants' RMS EMG activity for that stance phase (Grey band ± standard deviation). * = ST3 > ST1, ** = ST1 <ST2, *** = ST1 < ST5 for Adductor Group ($p<0.05$)

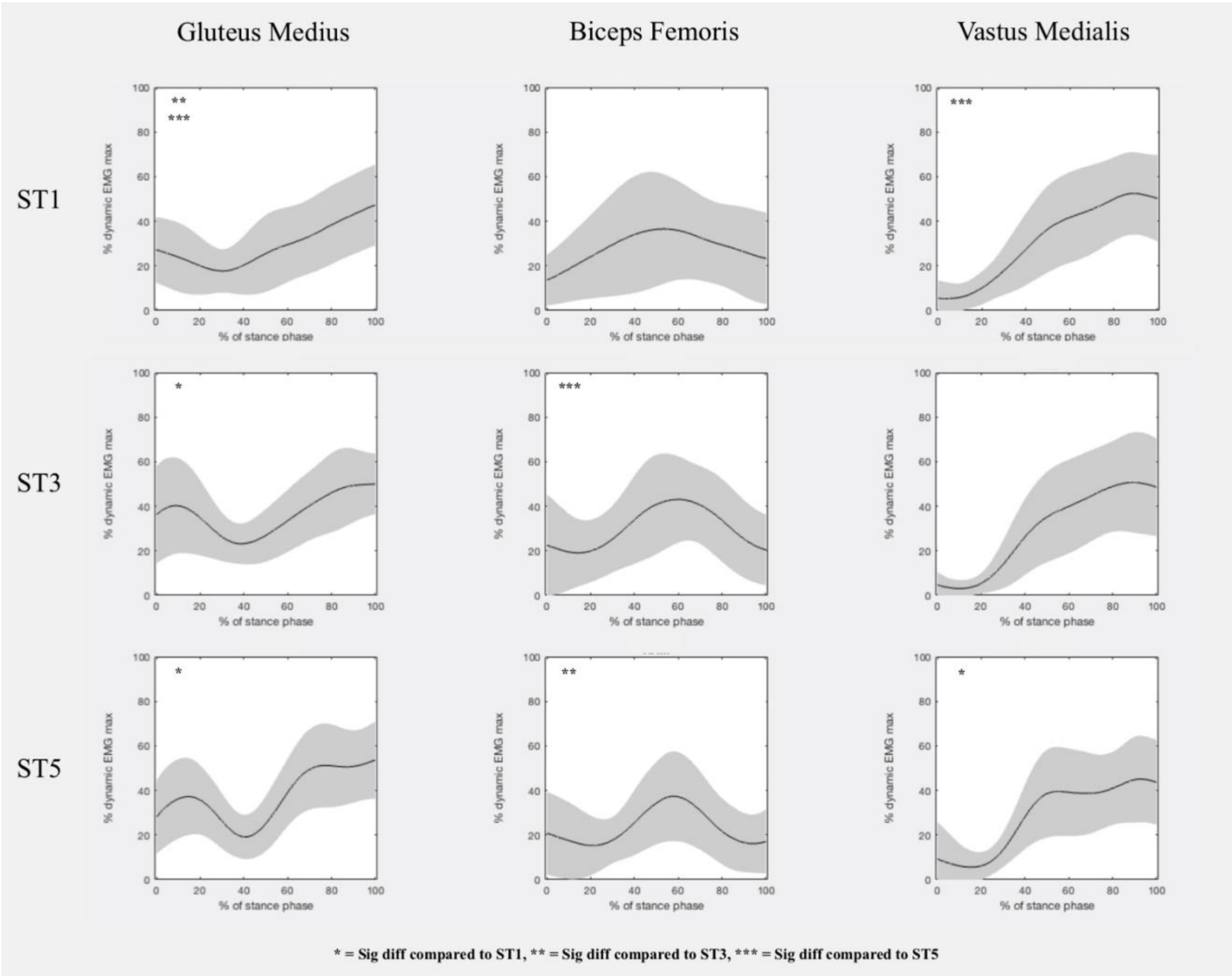

Figure 3b. Push-off stance phase Root Mean Square (RMS) electromyographic (EMG) activity waveforms (ST1, ST3, ST5) for Gluteus Medius, Biceps Femoris, Vastus Medialis muscles during a forward skating acceleration task.
Same notation as in Figure 3a.

3.3 Principal component analysis

The PCA analyses of the Gluteus Maximus and Biceps Femoris for the ST1 and ST5 phases allowed for the investigation of muscle activation variability during ‘acceleration’ push-offs as well as ‘steady-state’ push-offs. The first three *PCs* extracted in all four analyses accounted for over 90% of the EMG variability for the select phase (Table 6). For each *PC* extracted, the *PC-scores* (Z-scores) were compared between exotendon system conditions revealing no significant differences in likeness to the eigenvectors extracted (Table 7a, 7b). Visual inspection of *PC-score* clustering reflected the ANOVA results with a homogeneous mix of elastic conditions throughout the *PC x PC* plots. Participant *PC-scores* were much more distinctly gathered in all plots (Figure 4a-h).

Table 6. Percent variability explained by the first three *PCs* in four PCA analyses (ST1, ST5 x biceps femoris, gluteus maximus). A cumulative threshold of 90% variability explained determined the number of *PCs* to retain for investigation of the EMG activity waveform data.

Stance phase	Muscle	PC	% variability explained
ST1	Biceps femoris	1	66.7
		2	18.5
		3	10.7
		cumulative %:	95.9
	Gluteus maximus	1	74.2
		2	13.9
		3	6.6
		cumulative %:	94.7
ST5	Biceps femoris	1	57.4
		2	24.3
		3	8.9
		cumulative %:	90.6
	Gluteus maximus	1	71.4
		2	13.2
		3	8.7
		cumulative %:	93.3

Table 7a. *PC-scores* (±SD) for the first 3 *PCs* of stance phase 1 (ST1) grouped by exotendon condition (3 band; STAR, STAB, STAP, 1 control; STAC).

Muscle	Condition	PC1 Z-scores (±SD)		PC2 Z-scores (±SD)		PC3 Z-scores (±SD)	
Gluteus Maximus	STAC	20.5	(170.8)	-16.0	(72.6)	9.8	(23.0)
	STAR	6.1	(163.7)	21.6	(41.7)	3.2	(37.6)
	STAB	-10.4	(157.3)	2.2	(62.1)	-10.0	(23.5)
	STAP	-10.0	(167.3)	-7.2	(50.9)	0.1	(24.6)
Biceps Femoris	STAC	-5.7	(186.4)	-16.0	(55.4)	-3.2	(57.0)
	STAR	32.6	(177.4)	25.7	(77.1)	28.3	(50.0)
	STAB	2.5	(160.6)	21.3	(103.1)	1.1	(37.9)
	STAP	-36.8	(155.5)	-22.5	(73.2)	-19.8	(42.4)

Table 7b. *PC-scores* (±SD) for the first 3 *PCs* of stance phase 5 (ST5) grouped by exotendon condition (3 band; STAR, STAB, STAP, 1 control; STAC).

Muscle	Condition	PC1 Z-scores (±SD)		PC2 Z-scores (±SD)		PC3 Z-scores (±SD)	
Gluteus Maximus	STAC	15.2	(184.8)	0.4	(53.8)	4.4	(31.2)
	STAR	12.2	(149.4)	11.3	(62.2)	-4.0	(51.1)
	STAB	-8.1	(129.0)	0.5	(38.2)	4.7	(43.5)
	STAP	-10.0	(93.2)	-11.2	(43.0)	-10.4	(31.1)
Biceps Femoris	STAC	25.2	(167.1)	2.7	(82.5)	-18.4	(37.2)
	STAR	-51.2	(81.4)	2.2	(67.3)	1.4	(16.1)
	STAB	10.2	(88.1)	0.4	(82.6)	15.6	(40.0)
	STAP	3.6	(123.0)	-5.1	(86.6)	4.4	(40.8)

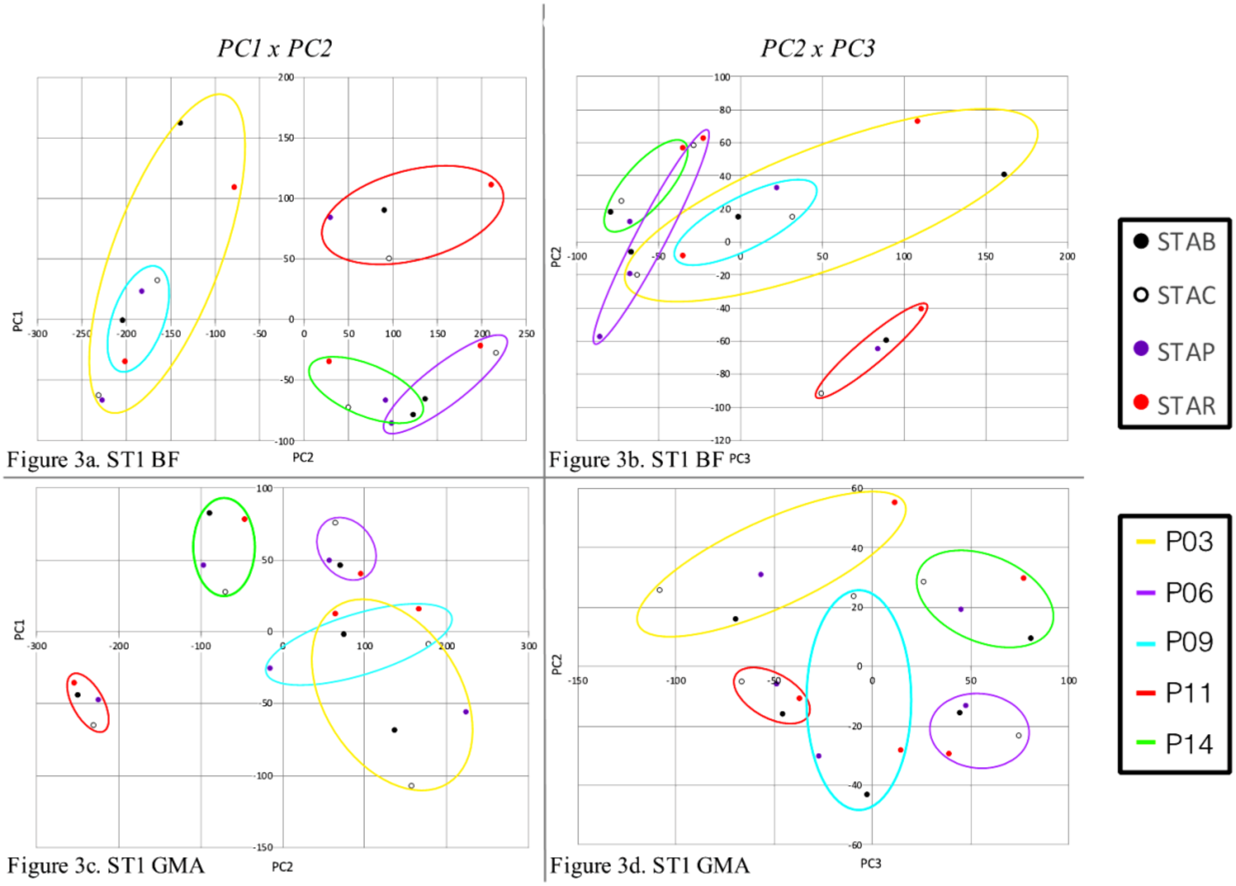

Figure 4a-d. Principal component by principal component plots (*PC x PC*) of the Biceps Femoris (BF) and Gluteus Maximus (GMA) principal component scores (*PC-scores*) during an 'acceleration' stance phase (ST1).
Dots were placed individually for every participant's exotendon elastic condition score (5 participants x 4 conditions; 20 dots total) relative to the *PCs* selected for the x and y axes (left subfigures = *PC1 x PC2*; right subfigures = *PC2 x PC3*).

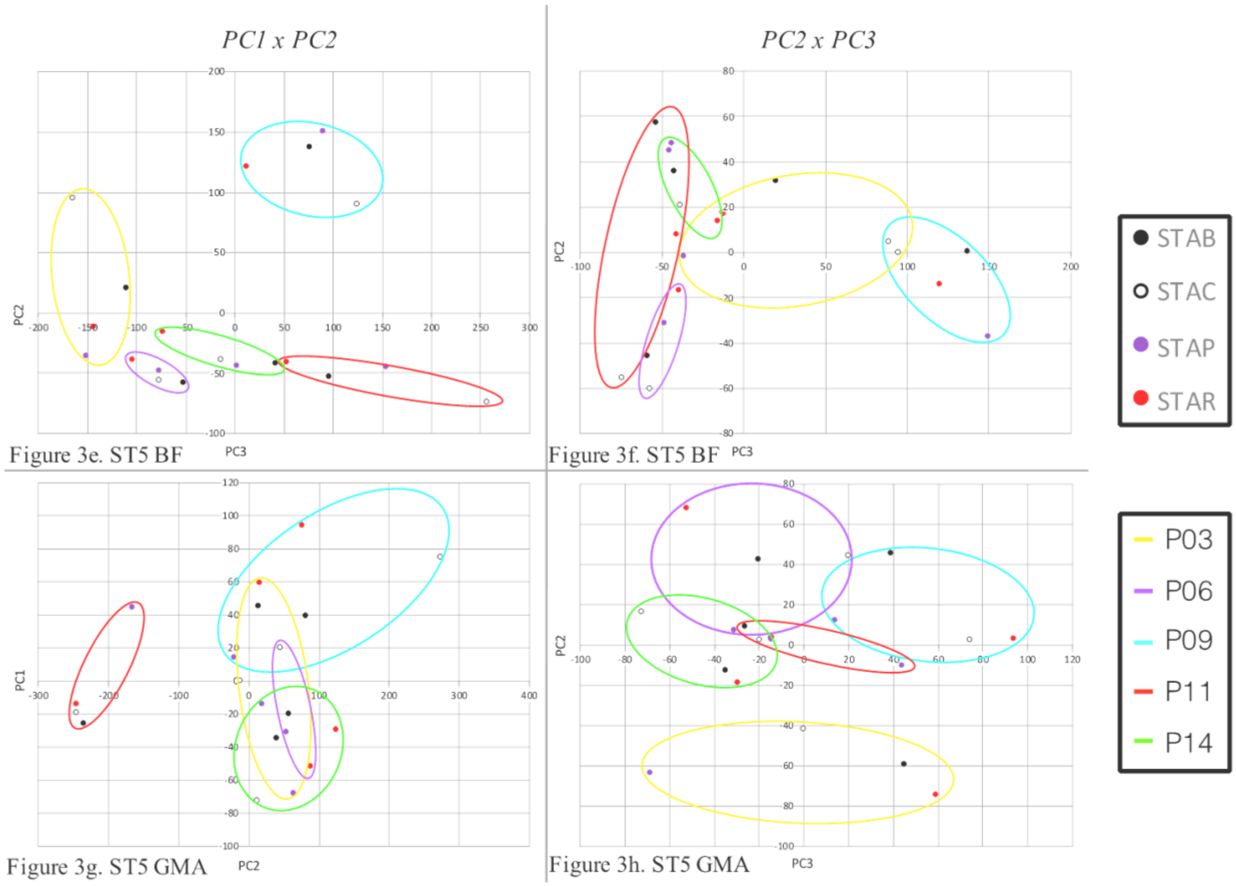

Figure 4e-f. Principal component by principal component plots (*PC x PC*) of the Biceps Femoris (BF) and Gluteus Maximus (GMA) principal component scores (*PC-scores*) during a 'steady state' stance phase (ST5).
Same notation as in Figure 4(a-d).

Despite the exotendon system demonstrating no substantial effect, the PCA analyses were able to flag important muscle activity waveform characteristics; magnitude, difference operators, or phase shifts (Brandon et al., 2013) (Figures 5-8).

For example, in ST1, reflecting an 'acceleration' glide and push-off, higher Gluteus Maximus *PC1-scores* showed greater muscle activation throughout majority of the phase (10-100%) (Figure 5a). For *PC2*, the higher scores showed a greater change in Gluteus Maximus activity towards the end of the phase (70-100%) when the push-off motion would have been occurring (Figure 5b). Higher *PC3-scores* demonstrated a Gluteus Maximus muscle activity difference operator as well, with a larger increase in the early stance phase (0-40%) where participant variability was highest, followed by a decrease in activity during mid phase (50-70%) (Figure 5c).

With regards to evaluating the Biceps Femoris activity during the ST1 phase, higher *PC1-scores* possessed greater muscle activation throughout the entire phase (0-100%) (Figure 6a). The higher *PC2-scores* demonstrated a steadier activation pattern with a smaller change in the late phase (70-100%) where muscle activity remained more elevated than the low score counterparts (Figure 6b). Muscle activity in *PC3* was more undulating with higher scores exhibiting lower activation during the early phase (0-40%) with a transition to higher activity during mid phase and the beginning of late phase (40-80%) and a drop at the end (80-100%) (Figure 6c).

PCA analysis of the Gluteus Maximus muscle during ST5, a ‘steady state’ stride glide and push-off, showed higher *PC1-scores* to be more activated from the end of early stance phase to push-off completion (30-100%) with a plateau in the increasing activity occurring during late phase (70-100%) (Figure 7a). Higher *PC2-scores* saw a lower change in activity with a more abrupt peak occurring from early to midstance phase (20-45%) during greatest participant variability, followed by a plateau and drop until the start of late stance phase (45-80%) (Figure 7b). In *PC3*, higher scores deviated from the rest with a plateau during the end of mid phase to start of late stance phase (55-75%) which then built up again to the end of the push-off (75-100%) (Figure 7c).

Lastly, *PC1-scores* of Biceps Femoris activity during ST5 showed higher scores exhibiting greater muscle activity for the entire longevity of the phase (0-100%) (Figure 8a). The higher *PC2-scores* showed overall less change for the waveform feature with a much higher starting muscle activity, that declined over the span of early stance phase (0-35%) before transitioning to an increase at the start of mid phase (35-50%) (Figure 8b). Finally, higher *PC3-scores* displayed greater change for the respective feature with slightly lower activity rising to a larger peak at the end of mid stance phase (55-70%), followed by a small decrease over late stance phase (70-100%) (Figure 8c).

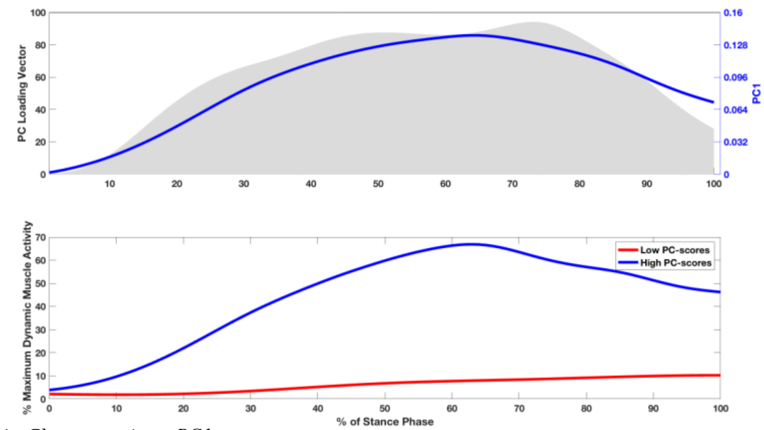

Figure 4a. Gluteus maximus *PC1*

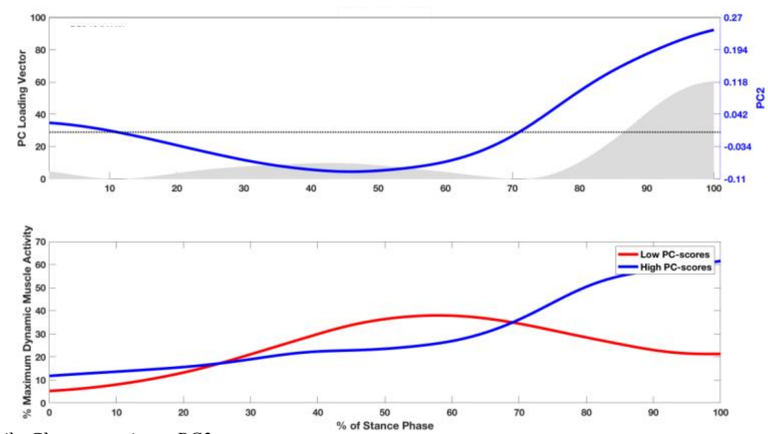

Figure 4b. Gluteus maximus *PC2*

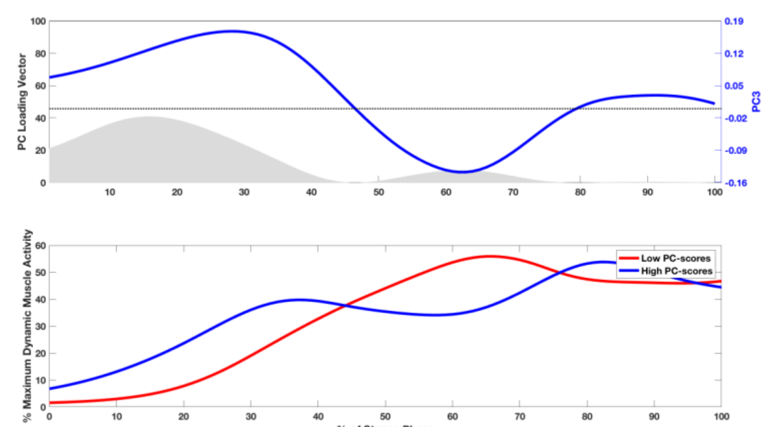

Figure 4c. Gluteus maximus *PC3*

Figure 5a-c. Gluteus maximus muscle activity principal component (*PC*) waveforms for *PC1* to *PC3* of an 'acceleration' stance phase (ST1).
PC loading vector; The top graph of each subfigure has the *PC* and *PC loading vector* plotted over the time (0-100%). The blue line represents the respective *PC* while the grey, filled waveform is the *PC loading vector*.
Singular *PC* reconstructions: The bottom graph in each subfigure represent a subset of participants with high (95th percentile) and low (5th percentile) *PC-scores* plotted as % maximum dynamic muscle activity over time (0-100%).

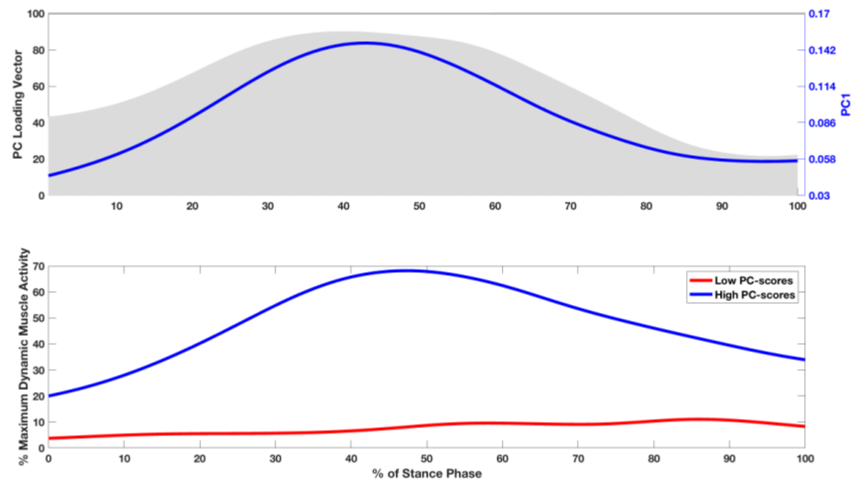

Figure 5a. Biceps femoris *PC1*

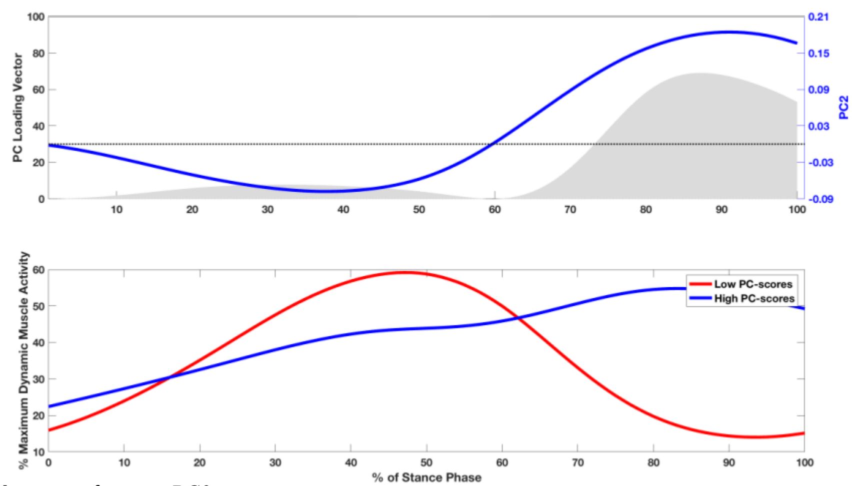

Figure 5b. Biceps femoris *PC2*

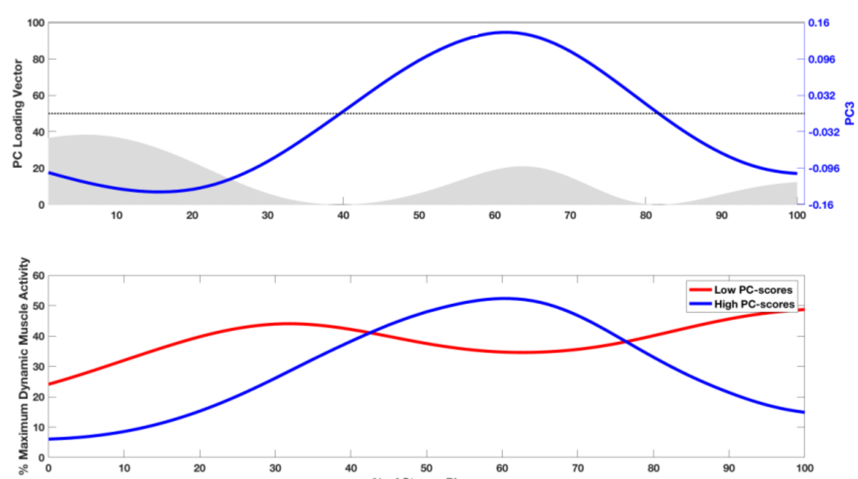

Figure 5c. Biceps femoris *PC3*

Figure 6a-c. Biceps femoris muscle activity principal component (*PC*) waveforms for *PC1* to *PC3* of an ‘acceleration’ stance phase (ST1).
Same notation as Figure 5(a-c).

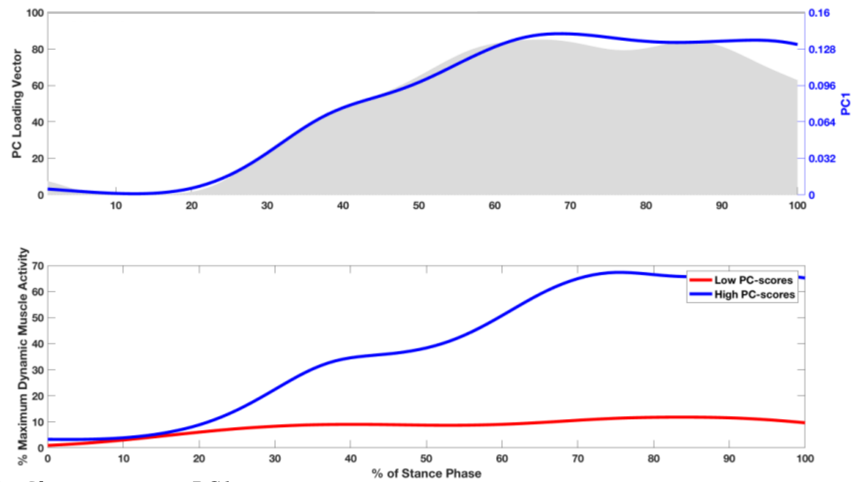

Figure 6a. Gluteus maximus *PC1*

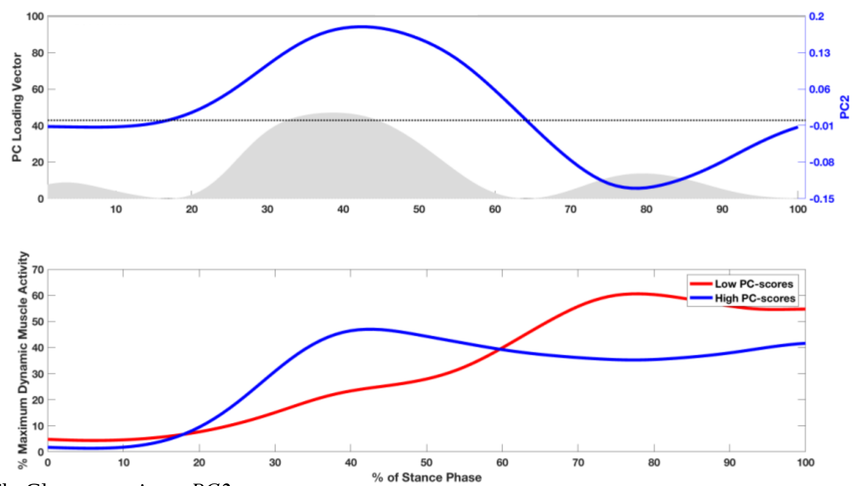

Figure 6b. Gluteus maximus *PC2*

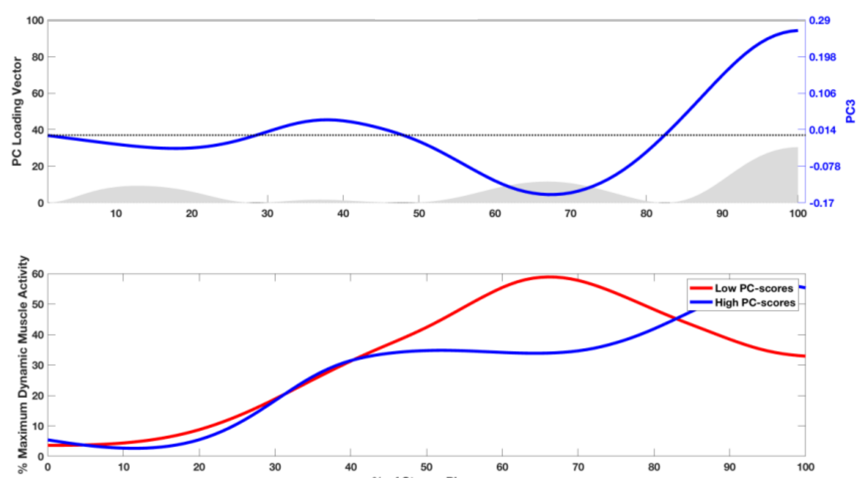

Figure 6c. Gluteus maximus *PC3*

Figure 7a-c. Gluteus maximus muscle activity principal component (*PC*) waveforms for *PC1* to *PC3* of a ‘steady state’ stance phase (ST5).
Same notation as Figure 5(a-c).

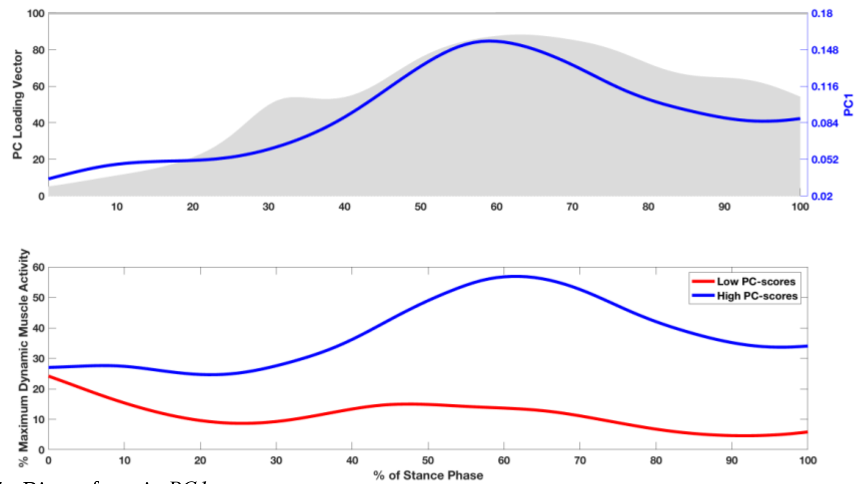

Figure 7a. Biceps femoris *PC1*

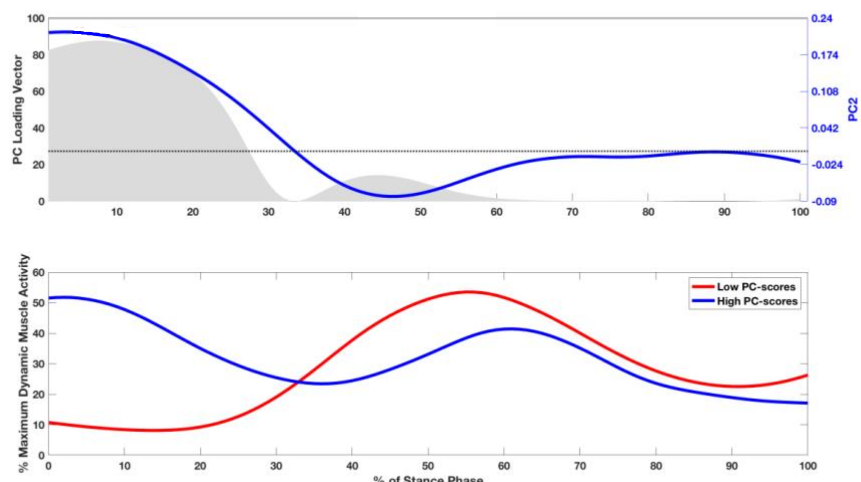

Figure 7b. Biceps femoris *PC2*

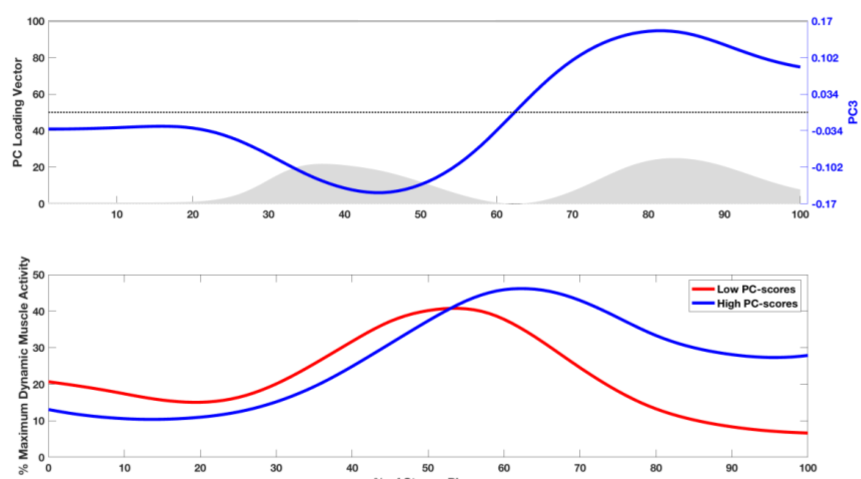

Figure 7c. Biceps femoris *PC3*

Figure 8a-c. Biceps femoris muscle activity principal component (*PC*) waveforms for *PC1* to *PC3* of a 'steady state' stance phase (ST5).
Same notation as Figure 5(a-c).

Chapter 4: Discussion

The passive exotendon system evaluated in this study was intended to improve skating velocity performance for forward skating accelerations. Specifically, employing elastic tension bands posterior to the hip joint may provide assistance with the eccentric-to--concentric hip extension muscles during stride push-off. However, findings from the previous hip kinematic study of the exotendon system revealed little gross changes in movement patterns during acceleration starts between tension band conditions and a control condition (McPhee, 2018), though participants noted greater perceived “power” and “stability” as well as fastest times for each individuals’ optimal tension band conditions. Hence, to better understand the sensitivity of lower body muscle activity to external passive exotendon assistance in hip extension, this study analyzed the activity of six leg muscles while wearing the exotendon system via wireless surface electromyography (EMG). The goal was to determine the effects of the bands on muscle activation patterns during forward accelerations, as well as stride to stride changes intrinsic to the skating start task. It is noted that due to the small sample size yielded in this study, the results must be interpreted with caution.

With our prototype passive hip extensor exotendon system, it was hypothesized that the tension bands would have perturbed hip joint muscle activity due to alterations in joint stiffness properties. Specifically, the system was designed to assist the hip joint in power production during forward skating (De Boer et al., 1987; de Koning et al., 1991). Potentially, hip extensor muscle activity could be reduced for the same amount of power production in accordance to the exotendons’ degree of stiffness. However, no differences in EMG root mean square (RMS) activity was seen throughout the stride between conditions nor a reduction hip extensor muscle activity between bands and the control. This indicated that the exotendon system, regardless of band stiffness, was not capable of provoking gross muscle activation changes nor assisting in reducing certain muscle requirements while attaining similar or faster task completion times. Changes in muscle activities across the task stance events did however occur as hypothesized from acceleration and steady state type strides, with a subset of the specific muscle predictions being support while others were challenged.

The anticipated changes from the hip exotendon suit was in thought feasible given prior skating studies that proposed body Center of Mass (CoM) dynamics where regulated through limb motion and stiffness, often discussed as the “spring-mass” concept in running biomechanics, and that CoM propulsion (both forward and vertical) was an integral trait of elite hockey players (Renaud et al., 2017). Many human hopping and running investigations allude to the idea of joint stiffness regulation as humans adapt limb stiffness for task execution (Grabowski & Herr, 2009). Mechanically speaking, it is suggested that overall task human movements under perturbation will remain predominantly alike which is in line with the pervious exotendon system findings. This maintenance of kinematics is thought to be accomplished by adaptations of the underlying musculoskeletal and neuromuscular components to maintain center of mass (CoM) dynamics (Buckeridge et al., 2015; Y. H. Chang, Roiz, & Auyang, 2008; Farley & González, 1996; Ferris et al., 2005; Grabowski & Herr, 2009). For example, it has been demonstrated that individuals decreased ankle stiffness through a reduction in plantar flexor muscle activity to accommodate a parallel ankle spring element (Ferris et al., 2005). In the current study, the hip extensor muscles did not mirror this conclusion, with no significant decrease in activity as response to the exotendon system elastic elements during skating. Furthermore, the absence of any alterations in lower body muscle activity challenges the concept of neuromuscular adaptation to joint stiffness altering equipment in powerful, dynamic sport scenarios.

One potential interpretation for this unexpected lack of muscle activity change alongside the vastly unperturbed hip kinematics may be that the bands physically did not possess adequate stiffness across the hip to alter lower limb biomechanics during maximal acceleration exertions. Whereas evidence has been found for changes in muscle activation as response to parallel a spring element at a smaller joint for a less dynamic activities (Farris & Sawicki, 2012; Ferris et al., 2005), it can be realized that this study's application of the exotendon system was potentially less akin to previous literature and consequently required more extreme elastic element properties to elicit quantifiable changes in elite hockey athletes. It should also be noted that the studies of ankle joint assistive equipment often site the role of the ankle as the primary regulator of leg stiffness in human gait, targeting it as an optimal site for evoking change in gait as well as its probability for capturing more exaggerated neuromuscular variability under altered leg conditions (Y. H. Chang et al., 2008; Farris & Sawicki, 2012; Ferris et al., 2005). As such, the insignificant findings for muscle activity stemming from a hip joint targeted system may also be result of joint selection for the exotendons, where provoked changes may have been systemically adapted for throughout the leg musculoskeletal structure during the absorption of loads and the transmission of power during skating.

Exotendon system aside, muscle activity was observed to change over the skate start interval as anticipated. It has been established in numerous kinematic hockey studies that individuals transition from a running-like stride (acceleration technique; ACC) to a more power-esque gliding stride (steady state technique; SS) over the course of forward skating accelerations from a stationary start (Buckeridge et al., 2015; McPhee, 2018; Renaud et al., 2017; Shell et al., 2017). As such, the alterations in limb kinematics must be accompanied by evolving muscle activity. A number of the muscles analyzed changed as expected over the acceleration task while some opposed the secondary hypotheses. Muscle activity waveform shapes (Appendix D) were found to align well with previous skating investigations employing EMG despite the statistical findings (de Koning et al., 1991) (Appendix E).

Hip extensor muscle activity, possessing a primary role in power production during forward skating, was found to support the non-exotendon predictions of consistent activation during push-offs while participants accelerated down the ice. The Gluteus Maximus muscle showed no change in activation between any of the push-off phases during the acceleration task, reflecting the findings of Buckeridge et al. (2015). This indicates the contribution of the Gluteus Maximus seems to be fairly constant, irrelevant of stride technique. In contrast, the Biceps Femoris muscle which contributes to both hip extension and knee flexion, decreased significantly from the middle acceleration task push-off to the late task push-off. This initially seemed contradictory to the hip extensor hypothesis, however the biarticular role of the muscle would have influenced the phase to phase comparisons. Lower knee extension linked to ACC technique may have been induced by more activated knee flexors like the Biceps Femoris, causing the drop in muscle activity during SS push-offs (Buckeridge et al., 2015; Shell et al., 2017). The nature of this muscle as biarticular across the hip and knee was likely the cause of the reduction in activity towards the end of the acceleration task.

In light of the greater knee flexor activity during ACC technique and greater knee extension previously identified in SS propulsion (Buckeridge et al., 2015; Lafontaine, 2007), the Vastus Medialis was thought to likely increase over the skating start task. This was not found to be the case with a significant decrease from the first push-off to the last. One plausible explanation for this occurrence could be linked to coactivation mechanics which is often used to comprehend joint stability (Frey-Law & Avin, 2013; Kellis, 1998) or joint kinematics during movement (de Koning et al., 1991). In line with this concept, the relative activation of agonist and antagonist muscles becomes important to interpret. The Vastus Medialis during the acceleration task should not only be examined in relation to the need for knee extension during propulsion, but also in regard to the knee extensor activity of the biceps femoris, and vice versa. It is known that antagonistic activity of knee flexors during knee extension works in joint stability to prevent anterior shear (Baratta et al., 2007; Kellis, 1998) which extends to all human motion including skating. With the decrease found in Biceps Femoris activity towards the end of the task, the Vastus Medialis would have needed to respond similarly in order to protect the knee joint despite an increase in activity to achieve high velocity extension (de Koning et al., 1991). The ratio of knee agonist (Vastus Medialis) to antagonist (Biceps Femoris) RMS activity was similar between the ACC and SS push-offs upon inspection with a slightly greater ratio during the later technique, which supports this theory in explaining the knee extension findings.

In relation to hip ab- and adduction, the Adductor Group, rarely investigated in skating, presented a rise in muscle activity from the early stance phase to the latter two stance phases in which push-off was occurring. This was most likely in response to greater eccentric activity as skaters have been found to increase hip abduction range of motion (ROM) and abduction velocity from ACC to SS strides (Buckeridge et al., 2015). Results from Renaud et al. (2017) would seem to counteract this explanation as hip abduction was discovered to be as equally prominent at the start of an acceleration task as the end. The acceleration task however in Renaud et al.'s (2017) study accounted for only three strides in total, suggesting participants may have still be transitioning from ACC to SS strides without ever attaining SS 'technique' unlike the number of strides accounted for in the Buckeridge et al. (2015) investigation. More important, the primary role of the Adductor Group was seen after push-off in the repositioning phase of the strides where activity was much more elevated. This logically aligns with the need to contract the adductors as means of bringing the skaters leg back under the body to accomplish the next stride. The identification of an increase in abduction from ACC to SS strides found during Buckeridge et al.'s (2015) investigation was also found to be in agreement with the Gluteus Medius findings from this study where muscle activity increased from the beginning to end of the acceleration task. Contraction of the Gluteus Medius supports hip abduction motion, aligning well with the increasing abduction ROM and the anticipated change in activation.

The sole ankle joint muscle captured, the Gastrocnemius, exhibited no change throughout the acceleration task in contradiction to what was expected. The work by Buckeridge et al. (2015) revealed greater Gastrocnemius activity during an ACC stride than an SS stride due to its biarticular functionality. The Gastrocnemius plays a primary role in contributing to ground reaction forces (GRFs) during running propulsion, which is commonly discussed as a parallel to early acceleration start technique. An increased muscle presence during ACC technique can be understood in this capacity. Its secondary action as a knee flexor has also supported the presence of greater muscle activity in ACC as Gastrocnemius co-contraction (along with Vastus Medialis) would help restrict (or stabilize) knee ROM during ACC (Buckeridge et al., 2015; de Koning et al., 1991). The greater knee extension kinematics previously found during SS would have suggested a reduction in knee flexor activity including that of the Gastrocnemius. Findings from the previous kinematic study did present a decrease in stride length and double support time when wearing the band system (McPhee, 2018) which would have masked normal skating conditions for these results. This may have caused participants to maintain the ACC technique the entire task, which would rationalize why we may have not seen a drop in Gastrocnemius activity towards the end.

The other counterintuitive finding involving the Gastrocnemius was that of greater activity during the repositioning phase of the stride following power production. One would assume that attainment of peak push-off velocity would require maximal plantarflexion as means of transitioning the kinetics through the body to the ice, however the present findings along with previous findings have indicated differently (de Koning et al., 1991; Lafontaine, 2007; Pearsall et al., 2001; Van Ingen Schenau et al., 1989). Explanation of this has been centered around the theory that skaters must maintain some amount of dorsiflexion until the skate has left the ice as to avoid undesirable catching of the blade tip that would produce extra friction. Plantarflexion is initiated during push-off prior to the blade coming off the ice however peak ankle plantarflexion velocity and ROM would be achieved in short succession. This aligns with the greater Gastrocnemius activity being captured during limb repositioning. In further justification of this finding, knee flexion, a secondary function of the muscle, is required for a player to bring their leg back underneath their center of mass to accomplish the next extension and push-off. This moment would contribute to increased gastrocnemius activity during the post propulsion phase.

The above interpretation stems from the conventional interpretation of muscle recruitment based on muscle RMS amplitudes. The alternate technique, Principal Component Analysis (PCA) provides insights into both amplitude and activation timing. PCA findings in relation to the effects of the exotendon system on muscle activity throughout the task supported the aforementioned theories behind the absence of interaction or differences being found from stride to stride. With focus retained to the hip extensors during both an ACC and SS push-off, muscle activity waveform variability was found to be unexplainable in accordance to the bands.

Specifically, band condition principal component scores (*PC-scores*) were found to be homogeneously dispersed across the most important features of variability being the first three principal components (*PCs*). This indicated that the bands did not significantly influence variability in muscle activation as they possessed similar amounts of resemblance to the *PCs*, lending evidence that either band stiffness was not enough to perturb the lower limbs in the applied scenario or the leg was capable of distributing any variability caused through numerous undetectable changes along the kinetic chain.

There was an interesting qualitative finding from the PCA however, lending to the possibility that the bands may affect muscle activity. When viewing the *PC-scores* plotted on *PC x PC* graphs, select participants seemed to have a particularly large spread across one or both features of variability. One explanation for this was that select individuals might be 'responders' to the exotendon system. This especially must be interpreted with caution as result of the small study sample size but potentially there exists a subset of the studied demographic that may be more susceptible to changes in stiffness at the hip in relation to an exotendon. The lack of significant differences in band condition *PC-scores* could therefore have occurred as result of suppression by system 'non-responders'. Inversely, it could be interpreted that those individuals with a greater spread of scores were unable to become accustom to the exotendon system within the time bearing the equipment, and that long familiarisation would have greatly improved clustering, reducing band effects. Ultimately a greater number of participants would be required to realize these conclusions. The statistical findings therefore must remain as the primary interpretation, meaning natural trial to trial variability is to be implied as a leading influencer in constructing the features of waveform variability.

PCA was subjectively compared between the ACC and SS strides to interpret how technique may have changed lower limb muscle activity within the elite skaters in this study. PCA of gluteus maximus muscle activity during the first push-off stance phase and the last push-off stance phase provided some novel insight into potential technique differences over the length of the acceleration task. For both ACC and SS stance phases the results identified muscle activity magnitude as the greatest feature of waveform variability. For the ACC stance phase, variation appeared almost immediately from blade to ice contact while for the SS technique, activity for the first fifth of the phase remained alike at a low percentage of maximal dynamic activation before becoming subject to large magnitude discrepancies. It can be suggested that this population of elite skaters seem to vary drastically with gluteus maximus contraction patterns from a stand-still position. Some individuals employing a more explosive activation of their hip extensors in reflection of a more rapid need for speed production while others utilize a steadier activation pattern to get moving. Once at higher speed however, a notable delay in initial muscle firing from an inactive state which seemed to be common across all participants before activity patterns began to increase variably.

The second most important feature of variability, *PC2*, displayed a much more unique gluteus maximus waveform pattern when comparing the ACC and SS push-offs. Both features of variation were that of difference operators, where there was a difference in muscle activity magnitude between local peaks and troughs (Brandon et al., 2013). For the ACC stride the differences in activation appeared at the end of the phase. Some participants were observed to have ramped up the contraction of the gluteus maximus in hip extension while others tapered off. In contrast during the SS technique, the variance in activation came earlier around the middle of the stance phase where a more abrupt increase and taper in activation was displayed by some while others presented a much slower, steadier ramping up in muscle activation. These differences suggested that skaters predominantly display rapid changes of activation during ACC just prior to the end of power production while this change occurred more consistently during early stance phase in SS.

The last feature of variability in the gluteus maximus waveforms was again identified as a difference operator. In the ACC stance phase participants initially exhibited larger variability beginning at the skate on event with activation patterns rising at a more rapid pace compared to other slower activators. Contrarily, the SS technique saw the change occur right at the end of the whole phase where a last prompt firing of the muscle occurred for some individuals during which others were decreasing. This suggested that participants underwent alterations in their muscle activation patterns at opposite ends of the stance phase depending on the type of stride being executed.

Application of the higher order PCA was also conducted for the biceps femoris muscle due to its direct implications in hip extension as targeted by the exotendon system. The first *PC* captured in the data was that of overall magnitude variability much like the gluteus maximus. For the biceps femoris however, differences in waveform magnitude was present from start to finish of both the ACC and SS techniques, inferring that an array of muscle recruitment patterns existed across the participant sample regardless of kinematic stride characteristics.

A difference operator presented as the second greatest feature of variability where larger discrepancies were identified in the final portion of the stance phase for the ACC stride while the opposite occurred for the SS stride with greater variation at the beginning. This suggested that a subset of the participant demographic was seeming to explosively activate their biceps femoris at the end of the strides early on in an acceleration task. In contrast, skaters seem to demonstrate prominent activation pattern variability at the beginning of the SS strides.

The final important feature of variability found in this analysis was another difference operator appearing in both ACC and SS stance phases. During ACC technique, activation differences could be distinguished right away in the early stance phase where activity was seen to increase at a faster rate for some compared to others until a number of participants peaked mid to late stance phase. The SS technique depicted changes occurring mid stance phase with some participant muscle activity suddenly dropping off while others continued to increase, followed by an amplitude difference in participant activity during the end of the phase. The differences in activation pattern changes between ACC and SS push-offs are primarily indicative of greater early variation within the first steps of an acceleration task compared to more evident variation in later stance phases.

A number of study limitations are apparent. Foremost and as frequently mentioned, sample size was the largest drawback in regard to the study's statistical power and subsequent strength during the interpretation of the results. A greater sample population of elite hockey players was required. The choice of lower body muscles from which EMG data was collected could have also been better tailored for an investigation of the exotendon system. Data acquisition from hip flexors would have been beneficial for the analysis. This however was not controllable as the data analyzed was collected prior to the inception of this study and muscle selection made in anticipation of a different experimental design. Finally, greater familiarization with the exotendons would have been optimal to better understand the source of muscle activity variability. A proper training protocol for the athletes to become accustom to the system would have been beneficial for result reliability.

To summarize, this study looked to examine the effects of a passive exotendon system on lower body muscle activity during maximal exertion forward skating accelerations in hockey. The findings were hoped to provide greater insight behind the slight time performance benefits observed by participants in the preceding kinematic study while wearing the system with their respective ‘best’ elastic stiffness. The band conditions displayed no differences in muscle activity across the task nor any effect on activation measure variability as found through PCA. Significant changes in muscle activity were found between stance phases over the task when ignoring the factor of the exotendon system. The condition findings would suggest that the chosen stiffnesses were resistive enough to evoke alterations in muscle activity patterns during a maximal forward acceleration task, even at the targeted hip joint. As such, greater resistivity from the exotendons may be required to elicit changes in skating biomechanics with the prospective outcome of more significant performance benefits. Further research is warranted to re-assess the concept of exotendons in hockey and to better verify current findings through a larger pool of athletes.

Chapter 5: Conclusion

This study investigated the activity of six lower body muscles during a forward skating acceleration task during the employment of a novel passive exotendon system, conceptualized with the goal of assisting extension at the hip to improve performance. Contrary to expectations, the bands were found to provoke no changes to normal muscle activity for the chosen task. This indicated that the elastic stiffness properties were not adequate enough to elicit alterations for the given scenario at the targeted joint. Principal component analysis (PCA), a novel form of analysis to EMG during ice skating sports, additionally returned no significant findings in relation to the exotendon system for the activity of two hip extensor muscles. A more thorough analysis of muscle activity across an entire acceleration task was accomplished through this study, linking past knowledge of lower body EMG during select single skating strides with a continuum of results while simultaneous offering new insight into the established evolution of skater kinematics over stand-still accelerations. The insight gained through PCA surrounding muscle activity variability within strides of a specific technique may also lend to identification of critical muscle patterns in athlete assessment, which could be used for the development of strength programs in attainment of performance enhancement.

Numerous future directions regarding exotendon implementation in ice hockey skating remain open. New directions for exotendons in hockey could prove valuable with implementation of elastics across more established 'susceptible' joints central to human movement such as the ankle. Collection of agonist-antagonist muscle pairings at targeted joints would also provide an interesting venue for analysis in regard to coactivation measures for interpreting joint moments or joint stability. This may catalyze an exploration of exotendons or assistive equipment in relation to injury prevention and rehabilitation, important for the longevity of athletes in the sport. Metabolic measurements alongside EMG would also prove valuable in future work to allow for calculations of metabolic power, a common measurement of exotendon success in human application. Ultimately, the current system design should be explored to a greater extent through an increase in elastic stiffness properties, investigation of skating tasks at submaximal efforts, or acute compared to prolonged exposure. Such modifications may uncover biomechanical changes linked to the system which remained eclipsed during the current hip exotendon studies.

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Appendices

Appendix A: Exotendon system design – Detailed description directly from the thesis by McPhee (2018)

"Development of the exoskeleton used in testing was guided through a Pugh matrix (Appendix D). This matrix showed that a soft exoskeleton would be more appropriate than a hard exoskeleton due to the relatively low cost, ease of adjustment and individuality associated with that type of system. The hip joint was chosen as the joint of focus as it is the primary force contributor to the skating stride, and could be incorporated in an area where participants are already accustomed to wearing restricting equipment. It was decided that the elastics would be secured to the participants using a system of straps rather than modifying pre-existing hockey equipment, as this reduced the potential of slippage, and increased the adjustability for each participant.

Early on in preparation for this study, numerous design choices were made regarding the construction of the system of elastic assistance which would be used in the testing. When deciding which joints the elastic elements should span, the entire lower extremities were considered. As the hip is the prime joint associated with skating power, it was selected as the area of focus for this system. The choice to use latex elastic bands came from their low associated cost, low weight, as well as high degree of adjustability. These attributes made them the clear choice when compared to rigid composite springs used in some other systems as shown through the Pugh matrix attached in appendix D. The system was also designed to be as light weight as possible. This meant minimizing the hardware required for the attachment of the latex elastics. A modified climbing harness (Momentum Harness, Black Diamond, U.S.A.), with aluminum cam buckles attached to the leg and hip bands was used. These cam buckles provided a secure hold on the latex elastics ensuring that no slipping occurred, while also keeping the weight down. When fully assembled, the entire system including latex elastics had a mass of only 850g. Velcro straps were sewn to the inside of the leg bands in order to ensure their placement on the legs did not change throughout testing. Through a full hip flexion range of motion, the system allowed for a change in length of the elastic elements of between 11 and 23cm depending on the size of the individual wearing it. Three different conditions of latex resistance elastics (Power Guidance Fitness Training, Suzhou, China) were selected for testing. Material properties of the latex were determined by measuring their change in length in response to added force in the method used by Triana and Fajardo (2012). The latex bands were hung from an aluminum cam buckle and weights were added to the free hanging end via a second aluminum cam buckle. Masses were increased in increments of 1kg until the length change in the elastic was greater than that seen in peak hip flexion. Five measurements were collected for each mass added."

Appendix D: Pugh Design Matrix for Elastic Assistance System Development

Matrix		**Cost**	**Ease of Adjustment**	**Individuality**	**Bulk**	**Number of possible variations**	**Total**
Exo type	***Soft***	+1	+1	+1	0	+1	4
	Hard	-1	-1	-1	-1	-1	-5
		Power contribution	**Supporting literature**	**Difficulty in design**	**Likelihood to get result**		
Joint focus	***Hip***	+2	+1	0	+1		4
	Knee	+1	+1	+1	+1		4
	Ankle	0	0	-1	0		-1
		comfort	**adjustment**	**individuality**	**Potential to slip**		
Attachment	***Equipment***	+2	-1	0	0		1
	Straps	-1	+1	+1	+1		3
		Time commitment	**Value of result**	**Enough by itself?**			
DV	***Time components (power) (acceleration)***	+1	+2	+1			+4
	Force	0	+1	0			+1
	EMG	0	+1	0			+1
	kinematics	-1	+1	0			0

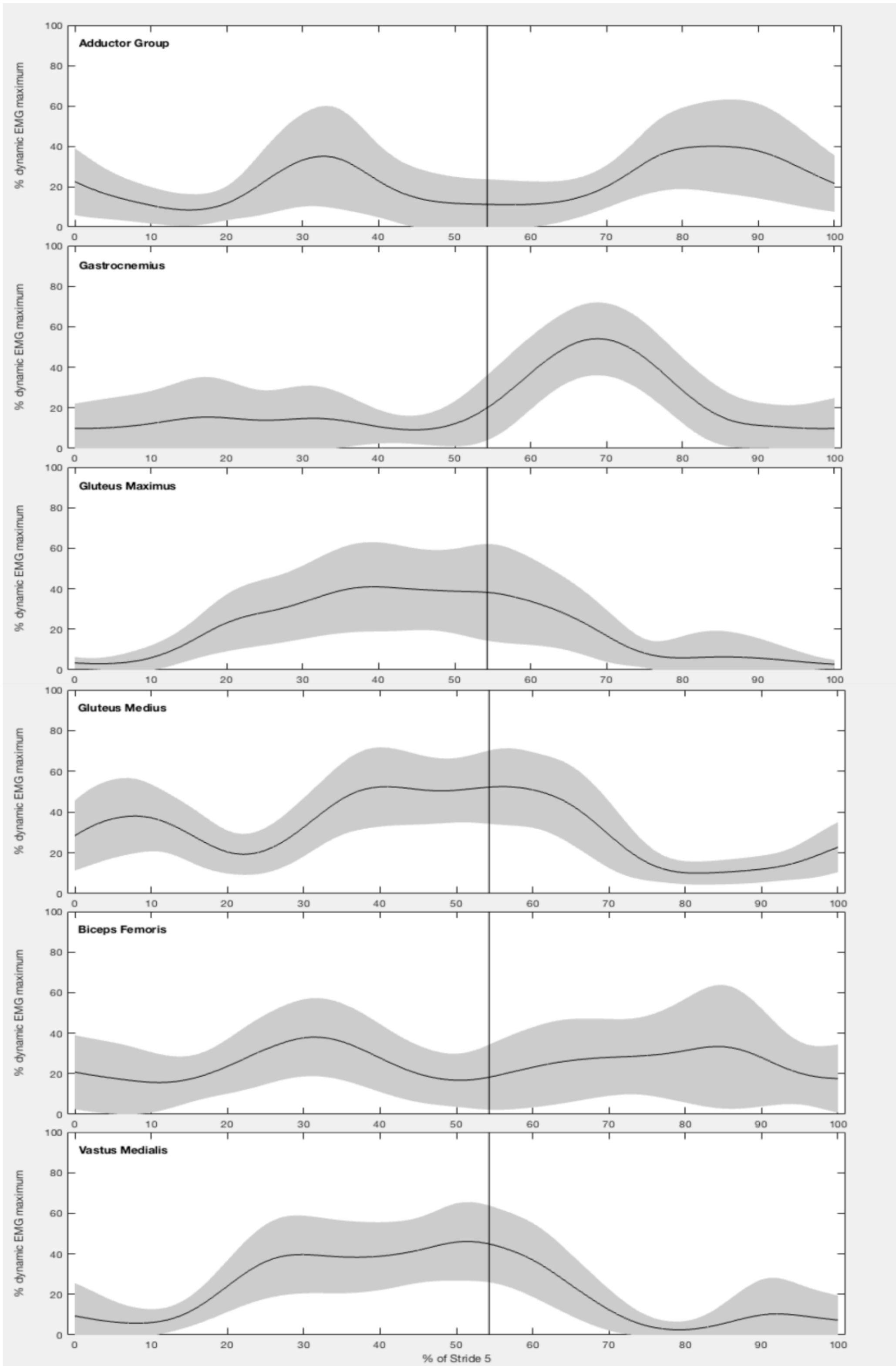
Appendix B: Exotendon material properties – Table directly from the thesis by McPhee (2018)

Elastic Condition	Cross Sectional Area (cm^2)	Average Stiffness (N/m) (±SD)
Soft	0.36	132.2 (20.8)
Medium	0.99	256.7 (38.1)
Stiff	1.53	440.8 (51.9)

Appendix C: Time performances for the acceleration task with each band – Table directly from the thesis by McPhee (2018)

Condition	Time to Skate 9 Meters (s) (±SD)	Best Condition	Worst Condition
Control	1.91 (0.01)	2	4
Soft	1.90 (0.08)	2	3
Medium	1.88 (0.08)	4	0
Stiff	1.92 (0.10)	1	2
Best Elastic	1.88 (0.08)	-	-

Appendix D: RMS EMG waveforms of 6 leg muscles from stride 5 of a forward skating acceleration in hockey. Muscle activity was normalized to a % dynamic EMG maximum value



Appendix E: Figure of mean leg muscle activity over a stride in speed skating – Figure directly from De Koning et al., 1991

Coordination in speed skating

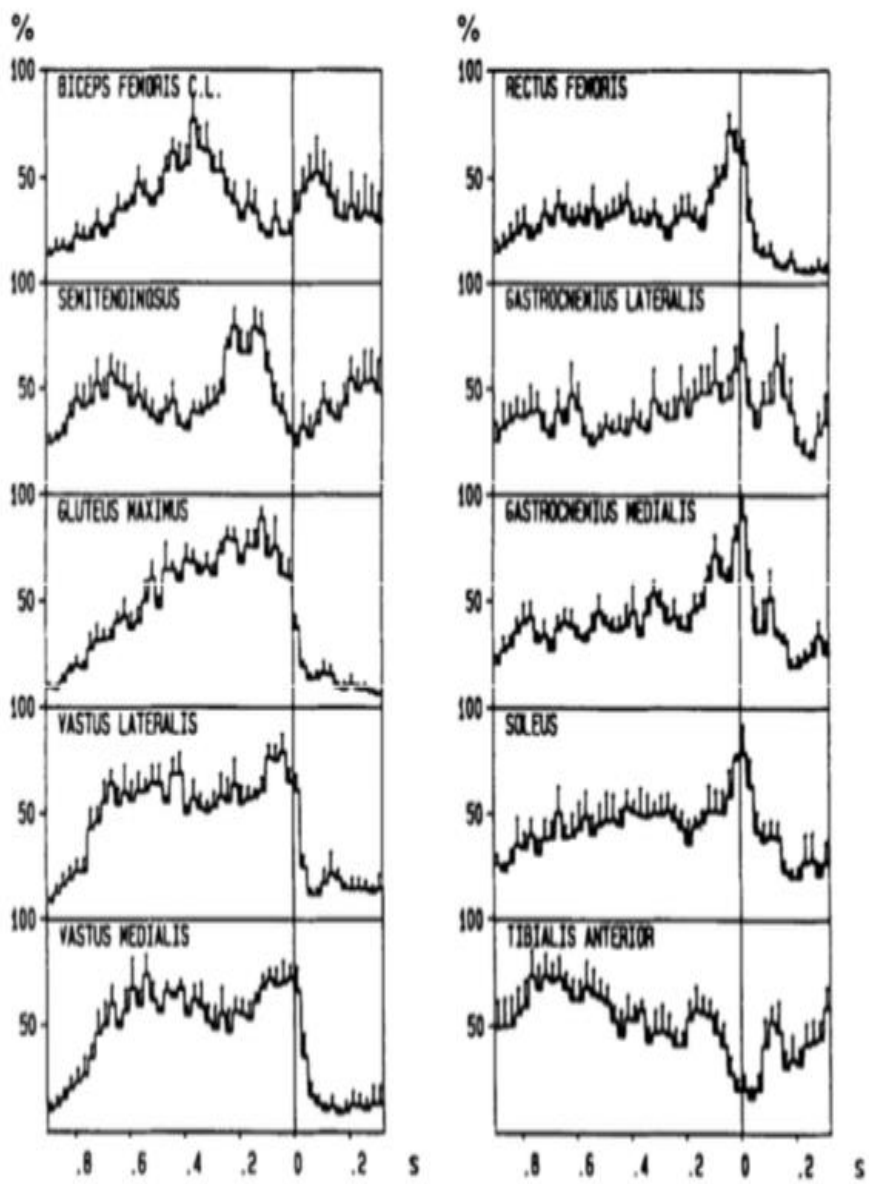


Fig. 6. Averaged EMG levels of the leg muscles of the elite subjects. Vertical bars indicate the standard deviations.