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Hamstring and calf muscle activation as a function of bodyweight support during treadmill running in ACL reconstructed athletes

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ABSTRACT

Rehabilitation after injury and reconstruction to the anterior cruciate ligament is thought to require a gradual reintroduction of loading, particularly during resumption of running. One strategy to achieve this is via the use of a reduced-gravity treadmill but it is unknown, if and how muscle activity varies in the reduced gravity conditions compared to regular treadmill running. Nineteen healthy participants and 18 male patients at the end of their rehabilitation (8 with a bone-patellar-bone graft, 10 with a hamstring graft) participated in this multi-muscle surface electromyography (sEMG) running study. The hamstrings and triceps surae were evaluated during a 16 km/h running while at 6 different relative bodyweight conditions from 50% (half weight-bearing) to 100% (full weight-bearing). Muscle activation was examined individually as well as normalized to a composite "entire" activation and considered across the entire gait cycle using Statistical Parametric Mapping. The healthy participants showed differences between the 50–100% BW and 60–100% conditions and in the hamstring graft group for 60–100% and 80–100% conditions. No differences were seen comparing all loading conditions in the bone-patellar-bone graft group. For the hamstrings, from 70% BW and above, there appear to be no difference in activation patterns for any of the groups.

The activation patterns of the hamstrings was essentially the same from 70% indicated bodyweight through to full weight bearing when running at 16 km/h. Accordingly, when running at this relatively high speed, we do not expect any adverse effects in terms of altered motor patterns during rehabilitation of these muscles.

1. Introduction

Following Anterior cruciate ligament reconstruction (ACLR), athletes are known to have deficits in strength [1], proprioception [2], gait parameters [3] and muscle function (e.g. quadriceps and hamstrings) [4]. However, there is no accepted "gold standard" rehabilitation protocol for how to best address these deficits [5]. Post-operative rehabilitation is thought to influence the ultimate outcome [5] with return to play for athletes post-ACLR being as low as 55% [6]. Long-term consequences such as increased incidence of arthritis are concerning [7] and while a graded return to sport is considered a cornerstone of rehabilitation programs following ACLR [8], there has been little research to objectively quantify this progression back to running.

Gradual re-introduction of running at full bodyweight post ACLR makes intuitive sense in an attempt to restore gait kinematics, facilitate muscle function, and decrease the incidence of anterior knee pain [9]. However, while it is known that Increasing running speed during overground and treadmill running [10] results in increased muscle activity by the athlete [11], little is known about the effects of altering bodyweight while running post ACLR.

One approach to reintroduce running loads gradually is via a weight supporting treadmill. More specifically, positive pressure, or "anti-gravity" treadmills allow creation of a controlled weight-bearing environment to train earlier in the recovery period [12]. Participants wear snug-fitting neoprene shorts sealed into an enclosed treadmill which has a load cell in the deck, under the belt. After calibration, the device can deliver an appropriate air pressure such that the participant is pushed away from the deck thereby reducing their effective body weight (BW) by the desired amount. The amount of reduction in BW commonly used can range from no reduction (100% BW) all the way down to 20% of BW. These devices have been used to study the effect of weight support on metabolic demand [13], joint forces [14] and plantar loading or vertical ground reaction force [15].

While reduced impact loading may allow for earlier resumption of running, it has been suggested that running at reduced BW may be associated with alterations in motor patterns [16] which, if adopted

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permanently, could be ultimately deleterious to an athlete's health once they resume "normal" running. Of particular interest in the sporting context is the possibility of muscle injury associated with running; the highest burden of which falls to the hamstring muscles [17]. The muscular work performed is shown to be non-linearly related to running speed with an increasingly higher load as speed increases [18]. Some research has focused on the EMG motor patterns during walking in a reduced gravity setting in healthy participants [19] and patients [20], but to our knowledge there are no studies focused on running in an ACLR patient cohort.

ACLR involves replacing the torn ACL and two common techniques used are the Bone-patellar tendon-bone (BTB) and the hamstring (HS) autograft ACL reconstruction [21]. Patients with an ACLR show altered movement patterns that may be related to the trauma caused by the surgery (to the hamstrings and quadriceps muscles respectively) and therefore diverse antalgic responses protecting the flexors/extensors are common. Since hamstring and calf muscle activation influence knee loading, it's of interest to examine these activation patterns which may effect risk for ACL re-injury.

Therefore, the aim of this study was to examine how muscle activation is altered during rehabilitation using different bodyweight in healthy male participants and two patient groups with BTB and HS graft after they have passed a criteria based rehabilitation program [5]. This paper focuses on muscle activation of the previously injured and non-injured leg to examine how each ACLR technique may affect the function of these muscles which are synergists of the ACL.

2. Methods

Control group: Nineteen injury-free male participants (age 35.4 ± 7.8 years, weight 77.6 ± 8.4 kg, height 179.1 ± 5.6 cm). Patients: Eighteen male participants who had undergone a recent ACL reconstruction and had completed all clinical criteria which allowed resumption of running. The patient group was subsequently divided into patients that underwent a HS graft (n = 10) (age 26 ± 3.84 years, weight 74.16 ± 7.19 kg, and height 176.89 ± 5.6 cm) or a BTB reconstruction (n = 8) (age 27 ± 7.69 years, weight 80.40 ± 9.44 kg, and height: 178.49 ± 7.29 cm). At the time of testing the patients were, on average, 7 months (± 2) post-surgery. Informed consent was obtained for each participant, and the experiment was conducted with the approval of the local ethics committee (ASPETAR: F2013000001). All participants volunteered to take part in this study and both patient groups were discharged from the rehabilitation facility and declared as ready to return to on-field sports specific activity.

EMG activity was recorded by means of surface electrodes from 6 muscles on both legs of the participant (i.e. 12 EMG channels). These included the plantar flexors soleus medialis (SM), soleus lateralis (SL), medial gastrocnemius (MG), lateral gastrocnemius (LG) and the hamstring muscles: medial and lateral hamstrings. The electrodes were placed by palpating the muscle bellies and orienting the electrodes along the main direction of the fibers [22].

The EMG sensors also contained 3D accelerometers, these inertial signals from the sensor placed on the shin in order to define the gait cycle during running [23]. All data was recorded at 2000 Hz using a Delsys Trigno Wireless System (Boston, MA), and filtered (high pass filter with 5 Hz cut off, low pass with 500 Hz cutoff). Heel strike was identified by minima in the shank vertical acceleration.

The running data was continuously monitored and each step was extracted and normalized from heel strike to heel strike leading to 101 separate time points (0–100%) (Fig. 1).

Six running trials were performed on an Alter G treadmill at a running speed of 16 km/h with varying BW support ranging from 50% BW up to 100% in 10% increments. The 100% BW running condition is the reference condition comparable to regular treadmill running without any support. To assess relative muscle activation across the different loading conditions, the EMG signal was normalized to its respective maximal contraction value obtained during the 100% BW running trials for each participant.

Trial conditions were presented in a random order for each participant. In each condition, after an initial familiarization and once participants reported feeling comfortable in the treadmill, measurements started. At this time the participants ran for a minimum of 6 (maximum 30) further strides. Data from all of these strides for each condition, for each participant were individually averaged for subsequent analysis.

Comparative EMG analysis was conducted for the entire time-dependent EMG signal using Statistical Parametric Mapping (SPM, version M.03) in Matlab (R2014a, The Mathworks Inc, Natick, MA). More typical statistical approaches are shown to be problematic when there are many comparisons made. SPM [24–27] is a method that allows analysis of time-series without reducing the dimensionality to only specific scalars (e.g. peak amplitude swing/stance phase). This approach was adopted to allow comparison of each 1% interval of the different running conditions without suffering from Type I errors (due to multiple comparisons). SPM and embedded vector-field analysis (multiple time-dependent muscle signals clustered within one vector) provides better control of type I and type II statistical error where multiple comparisons are conducted [28]. Clusters were formed based on the similar time-dependency and synergistic activation characteristics and to form time-dependent muscle activity vectors. We clustered the EMG-series of the SM and SL, the GM and GL and the medial and lateral hamstrings forming a Soleus, a Gastrocnemius and a Hamstring cluster, respectively.

The mean normalized muscle activity amplitude (per cluster and running cycle) and corresponding standard deviations at each of the 101 data frames is used for the statistical comparison of the entire time-dependent EMG signal (Fig. 1). To identify intra-group differences (left vs right leg, injured vs uninjured leg) of the multi-muscle EMG series paired Hotelling's T^2 statistics were performed. Following vector field analyses, post hoc tests were conducted on each vector component separately (paired T-statistic). The significance level was set to $p < 0.05$.

3. Results

The SPM vector field analysis of the cluster with the highest condition i.e. 50–100%; 60%–100%; 70–100%; 80%–100% and 90–100% are presented below (Fig. 2 and for more detailed presentation see the Supplementary data file).

Control: Differences were seen for the control group comparing the 50–100% BW and 60–100% conditions but no differences were seen above especially not in the Hamstring cluster.

BTB: No differences were seen comparing all loading conditions in the BTB group.

HS: Clear differences were seen for the hamstring autograft group comparing 50% up to 80% BW.

Above 80%, there appear to be no difference in activation patterns for any of the groups and clusters.

3.1. Increasing load dependencies

When comparing the cluster within the next higher condition i.e. 50–60%; 60%–70%; 70–80%; 80%–90% and 90–100% the SPM vector field analysis of the three clusters found no significant differences for all three cohorts between the six running conditions (Fig. 2).

4. Discussion

The aim of this study was to examine if relative loading influences muscle activation patterns during running in control and ACLR participants. When comparing the muscle activity of the triceps surae and hamstrings in this study, significantly different levels of normalized EMG activity were observed between the different loading conditions