ISB 2013 BRAZIL

XXIV CONGRESS OF THE INTERNATIONAL
SOCIETY OF BIOMECHANICS

XV BRAZILIAN CONGRESS
OF BIOMECHANICS

TREADMILL VERSUS OVERGROUND SOCCER-SPECIFIC FATIGUE: THE EFFECT ON HAMSTRING AND QUADRICEPS STRENGTH AND FRONTAL PLANE PEAK KNEE JOINT MOMENTS IN SIDE-CUTTING

Raja M Firhad Raja Azidin, Sean Sankey, Mark A. Robinson, Jos Vanrenterghem
School of Sport and Exercise Sciences, Liverpool John Moores University, UK
Email: R.M.Raja-Azidin@2011.ljmu.ac.uk

SUMMARY
A 45 min overground match simulating fatigue protocol imposed greater physiological responses, and a greater reduction in hamstring muscle strength than a treadmill simulation with similar average velocity, distance and activity profile, for male recreational soccer players. Knee joint loading during a side-cutting task did not increase for either of the simulations. A greater predisposition of anterior cruciate ligament (ACL) injury at the end of the first half and at the start of the second half in soccer matches for male players is likely explained by muscular imbalance rather than frontal plane knee loading.

INTRODUCTION
Several studies have attempted to investigate the effect of soccer-specific match simulation on markers of ACL injury risk, including muscular strength and balance [1,2,3] along with potential markers of injury during dynamic tasks such as side-cutting manoeuvres [4]. These studies have focused on different types of match simulations, either treadmill (TM) or overground (OG). However, no previous studies have investigated the differences induced by each type of simulation. This study aims to further the understanding of ACL injury risk by focusing on how treadmill versus overground simulations affect hamstring and quadriceps strength and knee joint moments when matched for velocity and distance covered.

METHODS
Fifteen recreational soccer players consented to participate in this study (age: 25.7 ± 4.3 yrs; body weight: 73.9 ± 8.0kg; height, 1.7 ± 0.1 m). In a single-group repeated measures design, participants completed twice a 45 min TM and twice an OG soccer-specific simulation. The OG simulation was modified from a simulation described by Small $et\ al.$ [4] and was designed to include multidirectional and utility movements, and frequent acceleration and deceleration. The TM simulation was then designed to cover similar distance and running velocity, yet involving no changes in direction and constant accelerations of 0.5 m·s^{-2} to transition between running speeds. Players were tested prior to exercise (time 0 min), at half time (time 45 min) and 15 min post-exercise (time 60 min). Tests existed of either five maximal dominant limb isokinetic contractions at $120°\cdot$s^{-1} for concentric quadriceps (Q_{con}) and eccentric hamstrings (H_{ecc}), or five trials of anticipated 45° side-cutting manoeuvres.

Each side-cut involved a sudden change of direction at 45° with the dominant foot landing on the force plate. To limit inter-trial variability, a successful trial was only valid if the approach speed was between 4-5.0 m.s^{-1}, and the foot landed entirely on the force plate. The order of the four sessions was randomized. Each participant had 46 spherical reflective markers positioned according to the LJMU-model, a 6-degrees-of-freedom eight segments model including feet, upper and lower legs, pelvis and trunk [5]. Functional hip joint centres and functional knee joint axes were calculated to maximally reduce error due to anatomical landmark placement over these joints [6]. Geometric volumes were used to represent segments based on cadaver segmental data. 3D kinematic data were simultaneously recorded using a 10 camera optoelectronic system sampling at 250 Hz. Knee joint moments were calculated using inverse dynamics.

Heart rate (HR) and rate of perceived exertion (RPE) were recorded every 5 min throughout the match simulations. From H_{ecc} and Q_{con} the so-called functional H_{ecc}:Q_{con} ratio was calculated. A two-way ANOVA with simulation and time as independent variables was used to identify significant differences, with $\alpha=0.05$.

RESULTS AND DISCUSSION
The HR and RPE responses were significantly greater during the OG simulation (HR 161.1 ± 10.0 beats·min^{-1}; RPE 14.4 ± 1.1) than during the TM simulation (HR 140.3 ± 15.0 beats·min^{-1}; RPE 12.2 ± 0.9), tending to increase as a function of exercise duration (see figure 1). These greater HR and RPE responses observed in the OG than the TM simulation may be due to its high accelerations, decelerations, and a broad range of utility movements. The close similarities observed during the present study with corresponding values reported from actual match-play support the adequacy of the OG simulation at replicating the demands of soccer match play.

The Q_{con} was not significantly changed at 45 or 60 min in both simulations. This suggests that the quadriceps musculature acts within its capacity for generating high forces for the duration of one playing half at least.
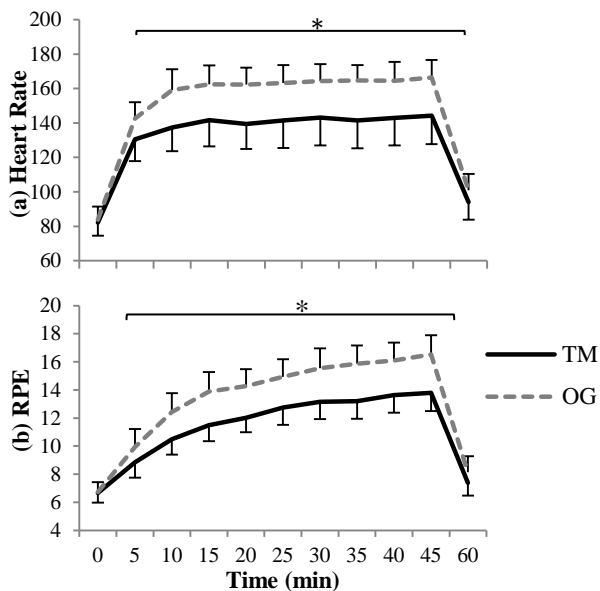
Figure 1: HR (a, beats.min^{-1}) and RPE (b) changes over time during treadmill (TM) and overground (OG) simulations. * Indicates a significant difference between simulations.

For H$_{ecc}$, however, a time effect was observed in that there was a reduction at 45 (12.7%) and 60 min (13%), but this was only the case in the OG simulation ($p< 0.01$, figure 2).
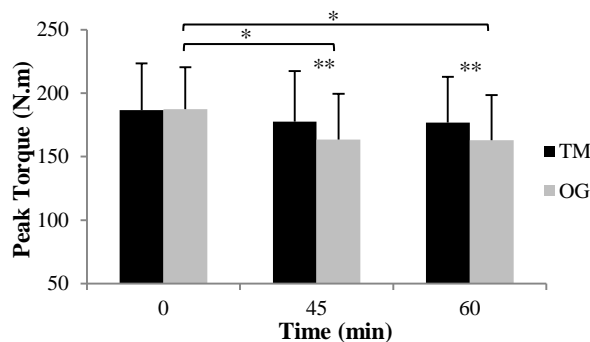

Figure 2: Eccentric H$_{ecc}$ peak torques during treadmill (TM) and overground (OG) simulations. ** Indicates a significant difference between simulations; * indicates a significant difference over time.

After 45 min there was a significant reduction in functional H$_{ecc}$:Q$_{con}$ ratio in both simulations (TM = 5.3% ; OG = 8.5%, $p<0.05$; Figure 3). Functional H$_{ecc}$:Q$_{con}$ ratio at the end of the passive half-time interval (time 60 min) was also significantly reduced relative to pre-exercise values in the OG (9.6%, $p<0.03$), but not in the TM simulation. The observed time dependent reduction in H$_{ecc}$ peak torques and functional H$_{ecc}$:Q$_{con}$ ratio with fatigue may help explain the reported increased predisposition to ACL injury during the last third of the first half of match play and during the early stages of the second half [7]. The OG simulation involves greater mechanical loading from greater accelerations and decelerations and the inclusion of utility movements compared to TM, so when monitoring player load for injury prevention these aspects should be taken into account, rather than only velocity or distance.

There were no significant time dependent changes in peak frontal plane knee joint moments for both simulations (Figure 4). This came at a surprise, as in females, an increase in frontal plane knee joint moments with fatigue had been reported [4]. This would indicate that males have an appropriate strategy for executing the side-cutting task without increasing loading on the knee joint. These are novel insights and further investigation of our data will hopefully reveal the nature of such compensation mechanisms. It is also still possible that unanticipated side-cutting or a longer soccer-specific simulation (90 min) may impose yet a greater challenge to the system and reveal increased joint loading after all.
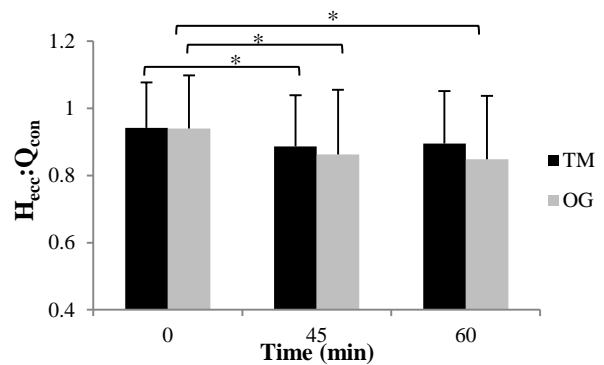

Figure 3: Functional H$_{ecc}$:Q$_{con}$ ratio during treadmill (TM) and overground (OG) simulations. *Indicates a significant difference over time.
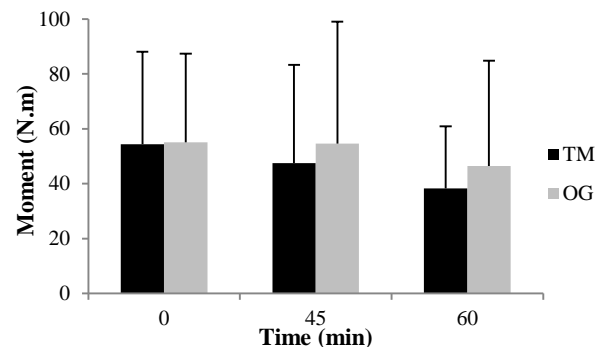

Figure 4: Peak knee abduction moment during side-cutting following treadmill (TM) and overground (OG) simulation.

CONCLUSIONS
For male players, a 45 min overground simulation imposed greater physiological responses, and greater reduction in hamstring muscle strength than a treadmill simulation with similar average velocity, distance and activity profile. However, knee joint loading was not increased for neither of the simulations. These results suggest a greater risk of ACL injury in male players at the end of the first half and at the start of second half in soccer matches due to muscular imbalance, whilst they are effective in keeping frontal plane knee loading low.

REFERENCES
1. Delextrat A, et al., *Int J Sports Med.* **31:** 192-7, 2010.
2. Greig M, *Am J Sports Med.* **36**: 1403-1409, 2008.
3. Small K., et al, *J Sci Med Sport*. **13**: 120-125, 2010.
4. Sanna G, *Clin. Biomech*. **23**: 946-954, 2008.
5. Vanrenterghem J, et al., *Gait Posture*. **31**: 517-21, 2010.
6. Robinson MA, et al., *J Biomech*. **26**: 1941-6, 2012.
7. Woods C, et al., *Br J Sports Med*. **38**: 36-41, 2004.