EMPIRICAL BASED MODELING APPROACH FOR THE QUANTIFICATION OF DYNAMIC KNEE STABILITY

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INTRODUCTION

Anterior cruciate ligament (ACL) injuries are common sports injuries that impact 1 in every 3,000 individuals [1]. An ACL injury results in the loss of dynamic knee stability, which is critical for the successful execution of movements like single-leg jump landing (SLJL) [2]. Discrete measures are often used to evaluate dynamic knee stability but these measures may not fully capture the altered joint biomechanics. To better capture the changing, we will implement an empirical based modeling (EBM) approach to quantify dynamic knee stability.

Previous work has assessed dynamic gait stability via methods such as; Lyapunov exponents [3, 4]. This method uses experimental (empirical) gait waveform data to derive models to evaluate gait pattern stability [3, 4]. The EBM approach employed here will also use experimental jump landing data to derive a transfer function to evaluate dynamic joint stability via bode stability analysis. A transfer function is ratio of the models' output response to the input perturbation [5]. Previous studies have used waveform data to develop transfer functions [6, 7]; however, they did not use the transfer function to assess stability.

Thus this study combined multiple techniques to evaluate dynamic knee stability from sagittal plane knee kinematics in athletes during a SLJL task [3-5, 8]. Surface electromyography (sEMG) data was also analyzed to determine how changes in joint stability relate to changes in muscle coordination. We hypothesize that this technique will be de able to differentiate between stable and unstable knee biomechanics and that greater knee flexor-extensor co-contraction will be exhibited during stable jump landings.

METHODS

Five Australian Football players (age 20 ± 1 yrs; mass 87.1 ± 5.4kg; height 1.90 ± 0.1m) were randomly selected from a larger cohort to performed six SLJLs each as part of the protocol (Fig. 1). Sagittal plane kinematics were obtained from experimental kinematic marker trajectories. And ground reaction force (GRF) and sEMG data were also synchronously recorded. The sEMG was collected for six muscles: medial and lateral vasti, hamstrings and gastrocnemii.

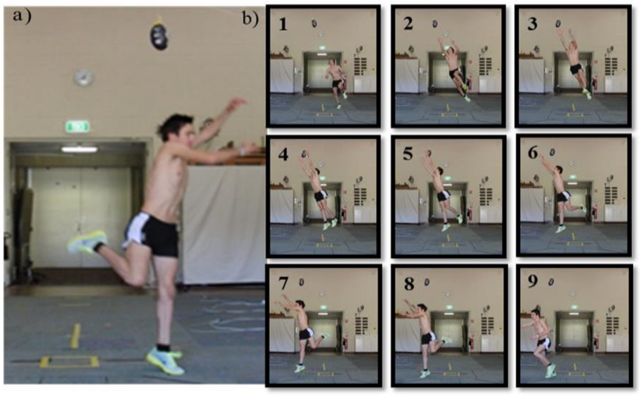

Figure 1: Series of images of an individual performing the experimental single-leg jump landing protocol in the laboratory.

The EBM transfer function was developed using the sagittal plane kinematic and GRF waveforms during the weight-acceptance (WA) phase of SLJL. Here the sagittal plane kinematic waveform represented the output response and the GRF waveform represented the input perturbation (Eq. 1).

$$Transfer\ Function = \frac{Sagittal\ plane\ knee\ kinematics}{Ground\ reaction\ force\ waveform}\quad Eq. 1$$

The Laplace transform converted the ratio from the time to frequency domain for the Bode analysis. The gain and phase margins obtained from the Bode stability analysis were used to quantify the individuals' dynamic knee stability. If both the gain margins (GM) and phase margins (PM) were positive, the individual was stable; otherwise they were unstable [5].

The sEMG data was normalized to each muscles peak activation. Directed co-contraction ratios (DCCR) were also calculated for the medial/lateral and flexion/extension muscle groups. The DCCR provides a value between -1 and 1 that indicates the role the agonist (flexor and medial) compared to the antagonist (extensor and lateral) muscles. We computed the integral of the sEMG data during the WA phase for the three muscles and the knee flexors (hamstrings and gastrocnemii) muscle group. A one-way ANOVA and Tukey's post hoc analyses determined if differences between the stable and unstable groups' PMs, sEMG DCCRs and area means were significant ($\alpha=0.05$).

RESULTS AND DISCUSSION

Twenty-one SLJL trials were classified as stable while nine were classified as unstable and the resulting stable and unstable GM and PM means were significantly different ($p<0.01$; $p=0.04$, respectively) (Table 1). The DCCR analysis found that individuals exhibited a more balanced knee flexor-extensor co-contraction during the stable trials than the unstable trials, although the difference was not significant. Overall, all three muscle groups exhibited greater muscle activation during the unstable trials than the stable trials (Table 1). Individually, the gastrocnemii muscles generated significantly greater activation during the unstable trials (85.7 ± 34.1) than the stable (57.8 ± 24.8) ($p=0.02$). Neither the quadriceps nor the hamstring increase in muscle activation were significant.

The results indicated that the EBM approach successfully assessed and quantified dynamic knee stability during the WA phase of SLJL. And that the stable landings exhibited greater knee flexor-extensor co-contraction. The gastrocnemii produced a significantly larger increase in activation during the unstable trials potentially to compensate for the smaller increase in muscle activation by the hamstrings. Previous research has shown that knee flexors and extensors function synergistically to stabilize the knee via joint compression [9]. The disproportionate increase in muscle activation exhibited by the muscle groups during the unstable SLJLs supports this concept. Furthermore, the larger increase in gastrocnemii muscle activation during the unstable landings could represent the muscle activation strategy individuals adopt in attempt to stabilize the knee during unstable SLJLs.

CONCLUSIONS

This study offers an alternate approach for evaluating dynamic knee stability. The results also provided insight the muscle activation strategies individuals adopt during stable and unstable SLJLs. Future work will investigate how muscle activation timing influences dynamic knee stability.

REFERENCES

1. Boden BP, et al. *Orthopedics*, **23**(6), 573-578, 2000.
2. Arden CL, et al. *Br J Sport Med* **48**(21), 1543-1552, 2014.
3. Dingwell JB, et al. *J Biomech Eng*, **129**(4), 586-593, 2007.
4. Stergiou N. *Innovative Analyses of Human Movement*, Human Kinetics, 2004.
5. Dorf RC et al. *Modern Control Systems*, Pearson Prentice Hall, 2008.
6. Gruber AH et al. *J Sport and Health Sci Solid*, **3**, 113-121, 2014.
7. Lafortune MA, et al. *J Biomech*, **28**, 113-117, 1995.
8. Morgan KD, et al. *J Biomech* **49**, 1686-1691, 2016.
9. Kvist J, et al. *Med Sci Sport Exerc*, **33**(7), 1063-1072., 2001.

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Table 1: Comparison of stability and surface electromyography area and directed co-contraction metrics for stable and unstable single-leg jump landing trials.

Analysis	Variable	Stable	Unstable	P-Value
Stability	Gain Margin (dB)	138.7 ± 75.7	-36.1 ± 41.3	<0.01*
	Phase Margin (deg)	67.8 ± 34.5	17.4 ± 96.5	0.04*
Area	Quadriceps	74.3 ± 17.0	91.1 ± 29.5	0.07
	Hamstrings	37.3 ± 18.5	56.4 ± 35.2	0.07
	Gastrocnemii	57.8 ± 24.8	85.7 ± 34.1	0.02*
	Flexors	95.1 ± 42.3	142.1 ± 69.1	0.03*
DCCR	Hamstrings/Vasti	-0.42 ± 0.25	-0.49 ± 0.21	0.45
	Gastrocnemii/Vasti	-0.08 ± 0.21	-0.24 ± 0.19	0.06
	Flexors/Vasti	0.28 ± 0.23	0.16 ± 0.21	0.17