TRAINING AND EXPERIENCE ATTENUATE PROTECTIVE GAIT STRATEGIES DURING BEAM WALKING

Michael Llewellyn, Ajay Bishop and Suzanne Scorer
DERA Centre for Human Sciences, Farnborough, GU14 0LX, United Kingdom

Walking on a narrow, raised beam is more difficult than walking across a floor. During beam walking, a protective strategy designed to maximise stability is adopted. This study compared the electrical activity (EMG) of selected leg muscles during normal walking with that during beam walking in novice and expert subjects. Results show that whilst changes (compared with normal walking) occurred in all subjects during beam walking, the magnitude of these changes is less in experts than in novices. In particular experts showed reduced muscle co-contraction during beam walking than novices. Thus whilst a protective strategy is elicited in expert subjects, the extent to which it is manifest is reduced. Experts maintain more typical patterns of EMG and should be less prone to muscle fatigue, a factor known to increase the risk of injury.

KEYWORDS: musculoskeletal Injury, balance, gymnasts, EMG, protective strategies

INTRODUCTION: During locomotion, if there exists a real or perceived threat of instability, the central nervous system (CNS) will adopt a protective gait strategy (Conrad *et al*., 1983). An example of a protective strategy is a response to encountering slippery or icy surface whilst walking. Such strategies prolong the periods of double-support and foot-flat contact and reduce the velocity of gait. The more difficult or complex the motor task, the lower the threshold at which the protective strategy is adopted. Protective strategies show features that suggest the CNS is maximising stability at the expense of velocity. During the development of gait from infancy to adulthood, a protective strategy is gradually superseded. The crawl of the infant is a protective gait strategy comprising a sequence of co-ordinated limb movements that is essentially quadrupedal. Crawling is stable, but is slow and metabolically inefficient. The more efficient bipedal, erect walk of the adult (which takes some 7 years to develop) is less stable and necessitates a fully developed CNS to provide the motor co-ordination necessary to maintain it (Popova, 1935). When the ability of the CNS to provide this control is compromised (for example in patients with certain sensory or motor deficits), the re-adoption of (the protective) essentially quadrupedal gait of infancy can often be observed. In healthy adults, the adoption of a protective gait strategy has clear benefits to minimise an immediate and temporary risk of falling. However, if maintained for extended periods, the muscle co-contraction that is typically associated with these strategies (Llewellyn *et al*., 1990) risks the onset of fatigue that could compromise the muscles' ability to generate and maintain appropriate levels of force. Under these circumstances the combination of instability and muscle fatigue may actually exacerbate injury risk. This study used a challenging motor task (beam walking) to elicit protective gait strategies in healthy adult subjects. Previous studies (Llewellyn *et al*., 1990; Bishop and Llewellyn, 2000) demonstrated that greater co-contraction of the *tibialis anterior* and *triceps surae* muscles occurs during beam walking compared with normal and treadmill walking. The researchers were interested in understanding whether, through training or experience, protective strategies employed by the CNS during beam walking, would be attenuated. This would be less metabolically demanding and reduce the probability of fatigue-related injury.

METHODS: The electromyogram (EMG) of 4 muscles (*tibialis anterior, soleus*, *flexor hallucis longus*, and *peroneus longus)* responsible for controlling foot and ankle movement during walking was recorded during level floor walking and walking along a 50-mm wide beam raised 600-mm above the floor. Data were gathered from 10 healthy female subjects. Three subjects were experts, being regional-level gymnasts. The remaining 7 were novices with no experience of gymnastics or kinaesthetic training. Subject details are shown in Table 1.

Table 1 Subject Characteristics

	Age in years mean (SD)	Height in meters mean (SD)	Motor dominance	Weight in kilograms mean (SD)
Experts (n=3)	20 (3)	1.84 (0.07)	2 R: 1 L	54.43 (7.06)
Novices (n=7)	20 (0.5)	1.65 (0.07)	6 R: 1L	63.04 (6.11)

Instrumentation and data collection: EMG was recorded using 4-pairs of gold surface electrodes (1 cm in diameter) attached to each subjects' dominant lower limb. Skin was abraded and cleaned prior to electrode attachment. EMG signals were amplified, band-pass filtered (F_c = 30-300Hz) and full-wave rectified using a 'leaky-integrator' with a 20ms time-constant (Neurolog system, Digitimer Ltd, UK).

In order to detect the following events: initial contact, toe off, stance and swing; and to calculate cadence, the plantar surface of the dominant foot was covered with conductive adhesive tape. The surface (floor or beam) on which the subject walked was also covered with conductive material. These two surfaces formed a switch which, when closed (by contact between the foot and the walking surface) generated a Transistor-Transistor Logic (TTL) pulse. When 'on', the pulse defined stance phase (with the rising and falling edges of the pulse provided the instants of initial contact and toe off). When 'off', the foot was in swing-phase.

Subjects walked along a 9-meter walkway traversing a force plate (Kistler 9281B, Alton, UK). In all cases the 600-mm long section of the walkway located directly above the force plate was mechanically isolated from adjoining structures (on the floor or beam) to ensure that forces would be recorded only when the foot impacted the force plate.

Subjects were investigated individually. Each subject made 10 replicates of each walking task (beam and normal walking). Data were only gathered from established gait. Thus, EMG data was typically gathered from 3-4 steps per walk, and force from a single step per walk. In total, for each walking task, between 20 and 30 step cycles containing only EMG and foot switch data, and 10 step cycles of force, EMG and foot switch data were recorded.

Data normalisation and analysis: Data were sampled at 1000 Hz per channel into a programmable interface (CED1401+ ADC, UK). Sampled data were examined to ensure that they were free from artefact. Data from each individual step (identified using the foot switch signal) from periods of established gait were selected and exported to a spreadsheet (Microsoft Excel). Within the spreadsheet data normalisation, averaging and graphing were performed as well as the calculation of temporal parameters.

Normalisation was carried out because intra- and inter-subject differences existed in the duration of each of the step cycles, and as a consequence they contained different numbers of samples. These differences must be removed before effective averaging can be performed. Normalisation was achieved using the technique of 'ensemble averaging' described by Winter (1983). This generates an average stride pattern of each of the variable across many repeat trials. The process of calculating the ensemble average was as follows. The start and end of the step cycles within each walking condition were identified based on foot-switch data (limb initial contact to next initial contact). For each step cycle the data samples within this period were interpolated to generate a time-normalised data set made up of 50 samples from the start (0%) to end (100%) of the step cycle. This provided a series of data sets, each representing a single step cycle and each of 50 samples, regardless of the duration, or number of data samples in the original data set. The mean and standard deviation were then calculated for each of these 50 normalised data samples (i.e., every 2% of the stride) and plotted over the stride period.

Averaged data were graphed and statistical analysis (one-way, repeated measures Analysis of Variance) used to determine significance, at the 95% confidence interval, of differences.

RESULTS AND DISCUSSION: Whilst all subjects showed changes in the temporal parameters of gait during beam walking compared with normal walking (Figure 1), those of the expert subjects were generally less affected than those of the novices.

Step duration (initial contact to initial contact) and the stance phase duration (initial contact to toe off) increased less in experts than in novices ($p<0.05$). There was no significant difference ($p>0.05$) between novices and experts in the extent to which stance to swing ratio and cadence (stride per minute) decreased during beam walking.

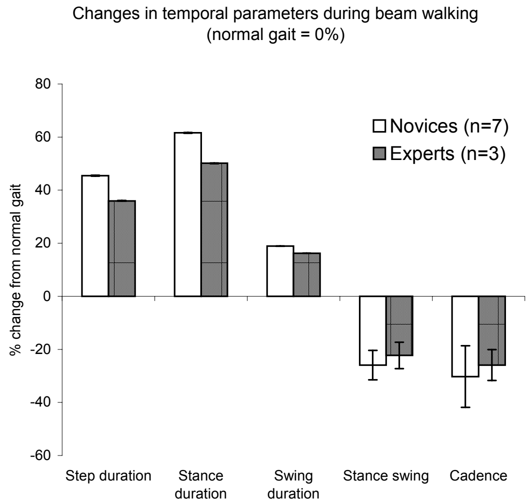

Figure 1 - Changes in temporal parameters of gait during beam walking compared with normal walking for novice and expert subjects.

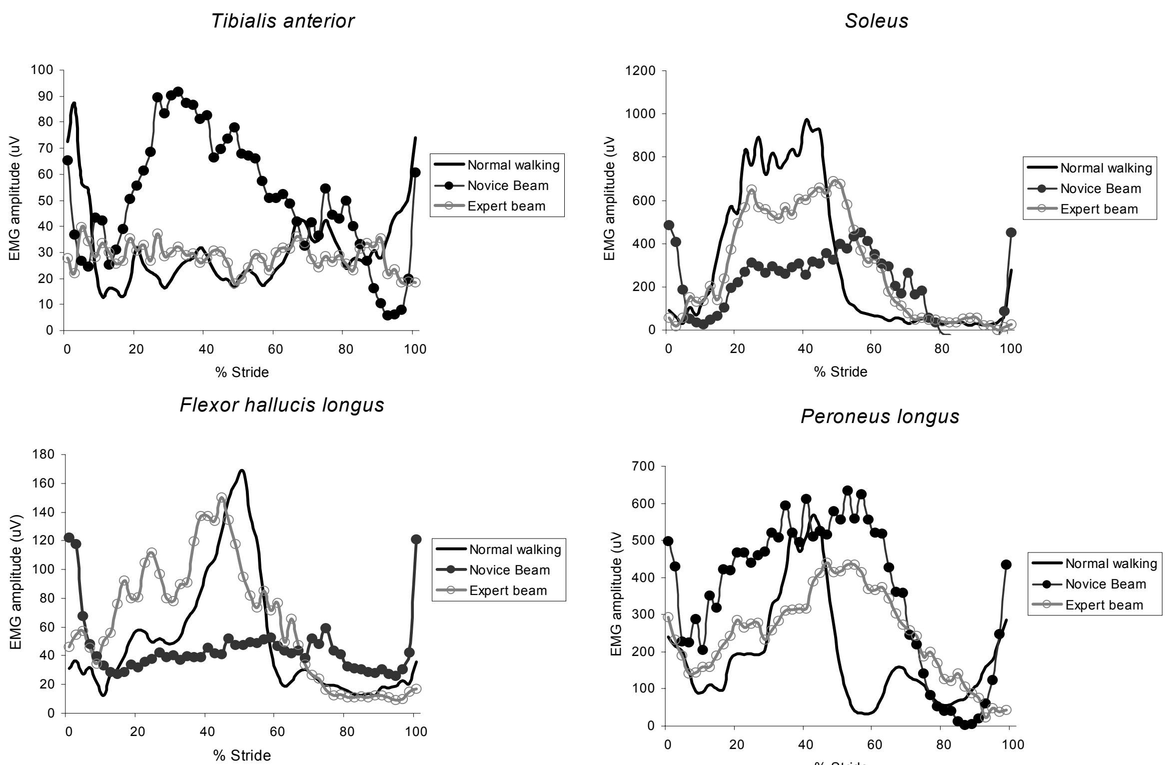

Figure 2 - Mean EMG activity during normal gait (n=10) and beam walking (experts n=3 and novices n=7).

Figure 2 shows normalised average time histories for the 4 muscles during normal gait and during beam walking. The data representing normal gait are from all 10 subjects (i.e., it comprises data from both the novices and experts). Data obtained during beam walking are from the specific populations (3 experts and 7 novices). For clarity the standard deviations calculated for each data point have not been shown. Comparing the ankle dorsiflexor *tibialis anterior*, with its plantar flexor antagonist *soleus* (Figure 2 top) it can be seen that, during stance phase (approximately the first 60% of the stride), beam walking produced changes in activity and timing in both novices and experts compared with activity during normal walking. These differences were however most marked in the novices. Novices showed consistent and high levels of tibialis anterior contraction during stance phase, a period when in normal walking the muscle is almost silent. Experts showed similar levels of tibialis anterior activation to walking, although the burst of EMG activity at the start and end of the stride normally associated with preparation for and control of ankle plantar flexion following initial contact were absent. During beam walking, the soleus activity of novices was reduced compared with that in normal gait and that recorded from beam walking experts. This suggests that the eccentric phase of soleus activation, during mid-late stance, was less pronounced and may reflect changes in the propulsive component of the GRFs.
In *flexor hallucis longus* and *peroneus longus*, during beam walking, novices showed greater differences in activity from normal walking than did the experts. The activity of *flexor hallucis longus* was much reduced during beam walking compared with normal walking in novices, whilst that of peroneus longus was much increased, almost as if compensating for the reduction in activity seen in the soleus muscle of these subjects. Taken together these results confirm that, whilst a protective strategy was seen in both expert and novice subjects performing a difficult balancing task, the functional consequences of the protective strategy are, to some extent, attenuated in expert subjects. As a result, they are able to maintain a more normal pattern of muscle activation and gait than novices.

CONCLUSION: Training and experience can affect the influence of CNS mechanisms that exist to protect the body during difficult motor tasks. By reducing muscle co-contraction associated with protective strategies it is likely that experts will have less risk of muscle fatigue than will novices. In turn, by influencing the way in which muscles are recruited during difficult tasks, training can provide some benefit in reducing the risk of injury arising from muscle fatigue.

REFERENCES:

Conrad, B., Benecke, R., Carnehl,J., Hohne, J. & Meinck, H.M. (1983). Pathophysiological aspects of human locomotion. In*:* J.E. Desmedt, (Ed), Motor control mechanisms in health & disease. (pp 717-726). New York, Raven Press.

Popova, T. (1935). Vsesoiuz.Issledovaniia po biodinamike locomotsii. In: N.A. Bernstein (Ed), Biodinamika khod'by normal'nogo vzroslogo muzhchiny. Chapter 3, Volume 1 . Moscow, Instit. Eksper. Med.

Llewellyn, M., Yang, J.F. & Prochazka, A., (1990). Human H-reflexes are smaller in difficult beam walking than in normal treadmill walking. Experimental Brain Research, 83: 22-28.

Winter D.A. (1983). The Biomechanics and motor control of human gait. Waterloo, Ontario, Canada: University of Waterloo Press.

Bishop A. & Llewellyn M. (2000). Muscle activity during beam walking in novice and expert subjects. In Y Hong & D.P. Johns (Eds.). Proceedings of XVIII International Symposium on Biomechanics in Sports (vol. 2, pp.943-944). Hong Kong: Department of Sports Science and Physical Education, The Chinese University of Hong Kong.

ACKNOWLEDGEMENTS: This work was carried out as part of Technology Group 5 (Human Sciences & Synthetic Environments) of the UK Ministry of Defence Corporate Research Programme.