

BIOMECHANICAL ADAPTATIONS FOLLOWING A LATERAL ANKLE SPRAIN INJURY: AN EXPLANATION FOR CHRONIC ANKLE INSTABILITY?

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The purpose of this study was to determine whether biomechanical adaptations play a clinically significant role in chronic ankle instability following lateral ankle sprain injury. Synchronised 3D motion analysis was conducted on 32 grade II lateral ankle sprain patients (grouped by functional stability score into copers and non-copers) during a dynamic cutting manoeuvre. Simultaneous EMG and force data were collected and compared for the injured and non-injured limbs. Copers could be distinguished from non-copers by certain EMG and ground reaction force parameters. Other distinctions could also be made between the injured and non-injured limbs. However these variables did not show significant group-by-side interactions to explain the symptoms of unilateral functional instability experienced by the non-coper group.

KEY WORDS: lateral ankle sprain, functional instability, kinematics, kinetics, emg.

INTRODUCTION: Chronic lateral ankle instability is likely to develop in 10-20% of patients following acute ligament rupture although incidences of up to 50% have also been reported (Barrett & Bilisko, 1995). The most common and debilitating form of ankle instability is functional instability (FI) and is normally described as an individual's sensation of 'giving way'. It is believed that clinical abnormalities, proprioceptive deficits, peroneal weakness, or mechanical instability are the primary causes of FI although these symptoms are not always evident in chronic sufferers. A possible cause that has not been well examined is the functional adaptation strategies adopted by the patient following injury. Analysis of the ACL-deficient population has revealed a subset of patients (copers) who are able to maintain high activity levels and experience neither instability, loss of function or weakness following ACL injury. Copers are able to stabilise the ACL-deficient knee during activity implying that their functional adaptations may be different from non-copers. Supporting this notion, research has shown differences in EMG (Sinkjaer & Arendt-Nielsen, 1991) and kinematic and kinetic parameters (Rudolph et al. 1998) between the two subgroups. As ankle ligament injuries are similar to ACL injuries in terms of the tissue types damaged and the mechanism of injury, similar adaptation strategies may explain the variation in functional stability following a lateral ankle sprain injury. If this is true, it is possible that adaptation strategies can be implemented or movement patterns modified post-injury to enhance the mechanical stability of the ankle joint and/or protect the injured ligaments. The purpose of this study was to investigate whether biomechanical differences existed between lateral ankle sprain patients of varying functional stability and to assess their clinical significance.

METHODS: Thirty two unilateral grade II lateral lateral ankle sprain patients (15M, 17F) participated in the study. To be included, all subjects must have had an ankle injury history of between 3-18 months, no other history of lower limb injuries or neuromuscular disorders, a plateau in recovery after which they had returned to sport, and be of recreational athlete status. Patients were grouped according to functional stability using an ankle score (14 copers (C), 18 non-copers (NC)) (Karlsson & Peterson, 1991) and interviewed about their injury rehabilitation. Every attempt was made to match the groups as closely as possible in terms of demographics and treatment history. Three dimensional movement analysis was conducted whilst 10 trials of a cutting manoeuvre (refer Figure 1) were performed in a laboratory setting. The cutting manoeuvre required the subject to land toe-first and in a slightly inverted position. The subject was asked to perform the task as quickly as possible and to maintain this testing speed for each trial. The testing order of limbs (injured (I) and non-injured (NI) was randomised. A modified Helen Hayes marker set (Kadaba et al. 1990) was used to model the lower limbs. The Helen Hayes marker set is a simple external marker set which was designed with a minimum of markers to simplify the identification of marker trajectories during gait. Simultaneous EMG data were recorded from the dynamic stabilisers

of the ankle (tibialis anterior, medial gastrocnemius and peroneus longus) using self-adhesive electrodes. Force data were collected under the landing foot using an embedded Bertec forceplate. Three-dimensional marker coordinate time histories were imported to Kintrak software for the calculation of joint angles, moments and powers. Customised labview software was used to analyse the EMG and force signals for temporal and amplitude characteristics. EMG and force data were collected and analysed in real time (ms) with respect to touchdown (TD). All kinematic and kinetic variables were presented normalised to stance. As chronic instability has been attributed to proprioceptive, strength and/or mechanical deficiencies, joint position sense (mean error in joint angle replication tasks), peak isokinetic eversion/inversion strength (peak torque) and passive range of motion (joint angle following application of 20 kg force) were measured using a Biodex isokinetic dynamometer. A 2x2 mixed ANOVA with a correction factor for repeated measures was used to compare group and side differences. Level of significance was set at $p < 0.05$.

RESULTS: The 2x2 ANOVA indicated that there were no significant differences in strength, proprioceptive error or passive range of motion between groups prior to testing. There were also no group differences in kinematic variables (refer to Table 1), however significant main effects by side were detected for ankle range of motion in inversion/eversion ($F(1,27) = 9.095$, $p = 0.006$, $ES = 0.55$) and plantarflexion/dorsiflexion ($F(1,27) = 7.952$, $p = 0.009$, $ES = 0.36$) with the injured limb of both groups demonstrating a smaller range of motion in each plane (refer Figure 2). The start time (refer Figure 3) and duration of EMG activity was not significantly different between groups or sides for any of the muscles investigated. Significant main effects by group were detected for the magnitude of peroneus longus activity at touchdown ($F(1,30) = 5.59$, $p = 0.025$, $ES = 0.61$) with the copers showing a greater magnitude of activity at TD respectively. Peak lateral GRF and loading rates were significantly different between groups – the copers demonstrated a significantly increased peak lateral ground reaction force ($F(1,30) = 7.36$, $p = 0.01$) and a significantly greater loading rate in both the lateral ($F(1,30) = 7.13$, $p = 0.012$) and vertical directions ($F(1,30) = 10.30$, $p = 0.003$) compared to the non-copers group. Significantly faster loading rates were also shown by the injured limb. There was no evidence of any interaction effects.

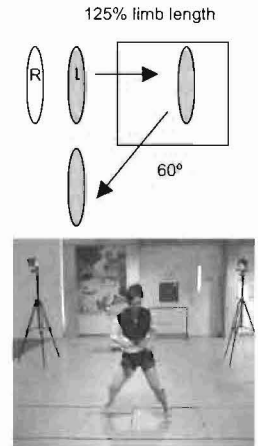


Figure 1. The cutting maneuver performed with the left foot landing on the force plate.

Table 1. Average (\pm SD) kinematic variables measured during the cutting maneuver.

VARIABLE	C,I	C,N	NC,I	NC,N
Ankle angle at touchdown ($^{\circ}$) (inv/ev)	9.2 (\pm 3.2)	11.5 (\pm 5.7)	12.1 (\pm 4.2)	11.9 (\pm 4.2)
Ankle angle at touchdown ($^{\circ}$) (pf/df)	-12.3 (\pm 7.7)	-14.2 (\pm 5.7)	-17.6 (\pm 6.4)	-16.4 (\pm 8.9)
Maximum angle ($^{\circ}$) (inv/ev)	5.8 (\pm 5.1)	2.7 (\pm 4.3)	5.1 (\pm 3.1)	3.6 (\pm 4.9)
Maximum angle ($^{\circ}$) (pf/df)	24.9 (\pm 5.3)	27.8 (\pm 4.2)	24.7 (\pm 3.2)	27.3 (\pm 4.2)
Range of motion ($^{\circ}$) (inv/ev)	-3.4 (\pm 3.7)	-8.8 (\pm 6.4)	-7.0 (\pm 4.8)	-8.4 (\pm 6.7)
Range of motion ($^{\circ}$) (pf/df)	37.2 (6.9)	42.0 (5.2)	42.3 (7.8)	43.7 (9.2)

DISCUSSION: The hypothesis that copers and non-copers would be distinguishable by kinematic variables was not supported. As most risk of injury and instability occurs with the foot in some degree of plantarflexion and inversion, (Singer et al. 1995) an inadvertent foot fixation at landing was expected in the symptomatic limb of the non-copers group. However, no differences in foot positioning were observed between group or between limbs at TD. There was a significantly reduced range of motion in the injured ankle in both the frontal (inversion/eversion) and sagittal planes (plantar/dorsiflexion) for both groups. This may suggest that both copers and non-copers stiffened the injured ankle during the cutting maneuver as a crude means of stabilising and protecting the previously injured ankle joint. The increased stiffness may have been partly achieved by increasing the amount of muscle

activity of the primary stabilisers about the ankle. However, the associated EMG data showed no evidence to support this or greater co-contraction about the injured ankle joint.

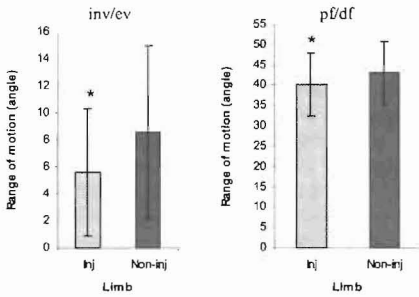


Figure 2. Average (\pm SD) range of motion from TD to peak angle in the inv/ev and pf/df planes. Significant differences are indicated ($p < 0.05$) (*)

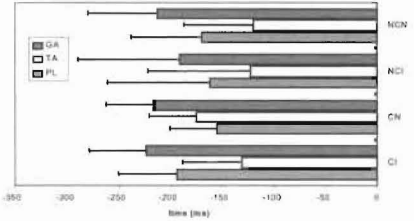
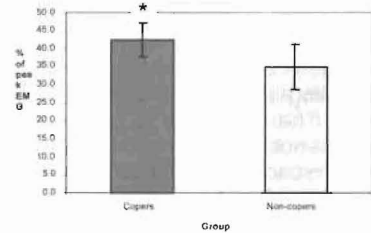


Figure 3. Average (\pm SD) muscle onset times for all group-by-limb combinations relative to TD (0ms).

Figure 4. Average (\pm SD) peroneus longus muscle activity at touchdown expressed relative to peak. Significant differences ($p < 0.05$) are indicated (*).



With respect to the EMG data, it was hypothesised that non-copers who suffer from repeated episodes of giving way may do so because they demonstrate a late onset of muscle activity (specifically the peroneus longus) which is of short duration and low amplitude. The EMG profiles showed that all three muscles were activated early in advance of ground contact and that there was no difference in onset time for any of the group-by-limb combinations. Therefore functional stability does not appear to be governed by EMG onset. The copers group exhibited a greater amplitude of peroneal longus activity at TD in both the injured and non-injured limbs. However, as EMG precedes muscle force development due to an electromechanical delay of approximately 20-100ms the actual muscle force associated with this signal does not develop until after TD. The contribution to functional stability at the vulnerable time of first contact is thus questionable. It was anticipated that the copers group would exhibit greater EMG activity, suggesting greater muscle activation and greater force production, in the muscles which act to evert and dorsiflex the ankle (peroneus longus) whilst the non-copers group were expected to show greater activity in the muscles which invert the ankle (tibialis anterior, gastrocnemius). However, there were no differences in overall EMG peak amplitude for any of the muscles investigated. Quantifying EMG amplitude has often been criticised as a number of factors can affect this (i.e. electrode configuration and cross-talk to name but a few). As these factors cannot be totally eliminated or controlled, normalisation of the EMG signal to maximum force and EMG amplitude is recommended when comparing between individuals or groups (De Luca, 1997). However, normalisation of the EMG amplitude to a maximal voluntary contraction (MVC) is difficult to achieve without the use of electrical stimulation. Therefore only the amplitude of the EMG signal was normalised and expressed relative to each individual's maximum EMG recorded during the cutting maneuver. With respect to the force data it was hypothesised that non-copers would move in a manner that would apply adverse forces and exacerbate symptoms of instability during lateral jumping movements. In contrast, non-copers appeared to modify their jumping technique so that they landed in a manner which exerted lower lateral forces at landing which would otherwise exert an inversion torque on the ankle joint. The slower loading rate (slope) also suggests that lateral forces were applied over a longer period of time in the non-copers group, and on the injured side of both groups. This may indicate that greater precaution was

taken by the less functionally stable group of patients in order to minimise stress on the ankle ligaments and that all patients jumped in a manner that protected the injured side. An alternative interpretation of the data would suggest that the copers group exhibited greater peak lateral ground reaction forces and faster loading rates as a reflection of their greater functional status and confidence on the injured ankle. Perhaps the most likely explanation for the lack of significant group findings is due to the variability of kinematic, kinetic and EMG profiles amongst patients. This may partly be attributed to the foot model used. This model differs to the Helen Hayes model which models the front foot and rear foot as one segment. As the front and rear foot can move independently during landing the toe markers were moved from the 2nd metatarsal to the navicular so that the movement of the rear foot was not masked. Subsequently, the front foot was not modelled and only the motion of the rear foot was analysed. This modified model has not been validated. The suitability of the marker set (which also uses protruding wands) during dynamic activities other than gait also warrants further investigation. Additionally, although every attempt was made to make the groups as homogeneous as possible, distinct variations in movement patterns were still evident amongst patients. These variations were still evident when the copers and non-copers were further divided into sub-groups (grouped by functional score, injury chronicity, and dominant limb).

CONCLUSIONS: This study demonstrated that copers and non-copers could be distinguished by certain EMG and force variables, but not kinematic variables during a lateral jumping task. It is possible that rehabilitation strategies should incorporate activities which aim to enhance the biomechanical differences observed. However the clinical significance of such practices remains uncertain as although group differences distinguished the copers from non-copers, the symptomatic ankle of the non-coper group could not be separated from other group-by-limb combinations to explain why symptoms were isolated to this ankle.

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