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Effects of reduced ankle dorsiflexion following lateral ligament sprain on temporal and spatial gait parameters

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Abstract

Partial rupture of the lateral ligament complex of the ankle is the commonest soft tissue injury affecting the lower limb. The effect of the limitation of motion at the ankle, particularly of dorsiflexion, on gait is unclear. In this study, 34 subjects were measured during their recovery from a partial rupture of the lateral ligament for both range of dorsiflexion and the temporal and spatial parameters of walking. Consistent relationships were identified between the range and the gait variables which were in concordance with the characteristics expected from an antalgic gait pattern. © 1999 Elsevier Science B.V. All rights reserved.

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1. Introduction

Ankle sprain is the commonest sports injury, particularly in activities involving running and jumping [1,2]. For example, partial rupture of the lateral ligament complex accounts for one injury per 10 000 of the population per day [3–5]. Jackson et al. [6]. reported that ankle sprains were the commonest injury at the United States Military Academy, West Point, with one-third of the cadets spraining an ankle during their 4 years there. The severity of the sprain can range from a mild injury, involving the disruption of only a few fibres of one of the ligaments forming the lateral ligament complex, to a complete rupture of the entire complex, with or without bony injury. The mechanism of injury is, however, typically one of inversion, plantarflexion and internal rotation, which will primarily disrupt the anterior talofibular ligament (ATFL) [7]. The patient with such an injury usually presents with pain and swelling over the lateral malleolus and adjacent hindfoot regions, with or without discernible bruising.

There is usually pain and limitation of plantarflexion and inversion, but the available range of dorsiflexion, too, may be substantially reduced [8], probably because of pain and swelling in the region. These restrictions can be expected to limit gait. The length of the stride will be restricted because the reduced ankle dorsiflexion constrains step length on the contralateral, uninjured side. Pain on weight-bearing decreases the period of single support on the affected side. While these manifestations of the ankle injury may be predicted in general terms, it is not known what relationships exist between the clinical measures of ankle motion and function in this condition. No previous study has attempted to follow the recovery of such an acute musculoskeletal injury with respect to the impairments manifest in the gait pattern.

The purpose of this investigation was to describe the changing characteristics of the gait pattern during recovery of ankle range of motion in patients following acute lateral ligament sprain. This particular in-

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jury is especially suitable for such an investigation, because it is normally a rapidly resolving condition in which there will be a concomitant increase in range of motion and decrease in pain.

2. Procedures

2.1. Subjects

Thirty-four (34) volunteer subjects were recruited from suitable clients presenting to the Physiotherapy Department of Calvary Hospital. Average age of the subjects was 26 years (range 15–48; S.D. 7.9). Equal numbers of subjects had sprained their left and right ankles (17 each). Twenty of the subjects had previously sustained an ankle sprain, and for 14 subjects this was their first ankle injury. No other neurological or musculoskeletal injury was reported in any subject. Participants were specifically screened to exclude any possibility of fracture at the ankle or injury to structures in the hind-foot region other than the lateral ligament. Subjects were attending for physiotherapy treatment and data were collected at each visit. Subjects gave informed consent before participating in the study. Ethical approval was granted by both Calvary Hospital Human Ethics Committee and by the Human Ethics Committee of the University of Sydney.

2.2. Measurement of passive ankle dorsiflexion range

A standard protocol was used, in which the subject's affected foot was positioned and secured by straps in a rig consisting of a footplate, hinged at a variable location to permit alignment with the ankle axis. The subject's shank was secured, using nylon straps, to a rigid frame. This device is known as the Lidcombe Template and has been demonstrated to have high intra- and inter-rater reliability [9]. The axis of the rig was formed by a gravity reference goniometer, calibrated to measure position from a neutral reference position, corresponding to plantigrade. The subject was instructed to relax and not to offer any assistance or resistance to the movement. The footplate was moved from a plantarflexed position into dorsiflexion by the researcher, using a handle incorporating a force transducer[1] until the subject experienced pain, or until a force of 100 N was applied to the footplate (corresponding to a torque at the ankle of 12 N m). The angle of dorsiflexion achieved at each visit was measured from the goniometer.

[1] Xtran 450N, Applied Measurement, Oakleigh, Victoria, Australia.

2.3. Derivation of temporospatial parameters of gait

Gait was recorded using a conventional video-camera and recorder system[2]. The camera was located perpendicular to a level walkway of total length 7 m. Subjects, dressed in shorts and wearing no shoes, were filmed at a shutter speed of 2 ms as they walked along the walkway. Adhesive markers were attached to the feet to improve visualisation. The field of view of the camera was the central 2 m of the walkway, an arrangement which provided a resolution to approximately 1 mm and ensured that subjects were walking at 'steady-state' speed when data were captured. The camera image was calibrated against a rigid frame of known dimensions. The videotapes of the subjects' walking patterns were over-dubbed with a time code with a resolution of 0.01 s. Thus, on field-by-field playback, sensitivity to 20 ms was obtained. Initial video playback was used to derive temporal events [10,11]. The times of initial foot ground contact and loss of foot contact on each side were recorded for each trial in turn. From these events double and single support times for the affected and unaffected sides and the total stride time were calculated. Double support was defined as the period between the commencement of foot contact on one side and the loss of foot contact on the other side. In the case of these subjects, limitation of ankle motion meant that, in the early trials particularly, the foot contact was not necessarily represented by a clear heel strike. Single support was defined as the period between loss of foot contact on the contralateral side and commencement of the next contralateral foot contact. Stride time was the time from one foot contact to the next initiation of foot contact on the same side.

The spatial coordinates of the foot markers were realised using a manual digitising tablet and the video-playback unit in a configuration similar to that described by Abraham [12]. Images from the video recorder[3] and from an overhead camera which captured the active area of a digitising tablet[4] were mixed[5] and displayed on a monitor. Prior to each test, a calibration frame of known dimensions (1 m^2) was filmed in the centre of the target zone. The video film was advanced or rewound until foot location during each stance phase of the cycle could be clearly identified. The digitising tablet has a theoretical resolution of less than 0.1 mm. The coordinates realised were stored using commercial software[6] and subsequently analysed in a customised

[2] National Panasonic, Panasonic Inc., Secaucus, NJ.

[3] National AG6200 freeze-frame recorder, Panasonic Inc., Secaucus, NJ.

[4] SummaSketch II Professional Plus, Summagraphics Inc.

[5] National WJ-SIN, Panasonic Inc., Secaucus, NJ.

[6] SigmaScan, Jandel Scientific Inc.

program developed in-house. This program computed stride parameters according to conventional definitions [13]. Thus on completion of the process, data were derived for stride and left and right step lengths as well as the temporal variables described above. In addition, stride velocity was calculated from the formula:

Stride velocity = *stride length*/*stride time*

Seven walks were filmed at each visit.

2.4. Data analysis

Data derived from each set of seven walks were averaged. Subjects visited the Physiotherapy Department on a variable number of occasions, recordings of gait and ankle range being taken on each occasion, giving a total of 239 observations. Correlational analysis was performed between the key variables of interest (stride velocity, cadence, uninjured side step length and ankle range of motion; single support time on the injured side and stride velocity) and between the other temporal and spatial variables.

Symmetry was calculated for step length and for double and single support times using the formula [14]

Symmetry index = *Affected side*/[*Affected side* + *Unaffected side*]

Perfect symmetry is expressed by an index of 0.5. Other temporal relationships were expressed as a percentage of total cycle time. These data were grouped together according to ankle dorsiflexion angle, with each cluster falling within 2° of ankle motion. Analysis of the symmetry indices and principal temporal phases (braking and thrusting double support, single support and swing) was conducted using Kruskall–Wallis one-way analysis of variance. This analysis tested the hypothesis that no significant difference would be found between these temporal and spatial variables ordered by ankle range of motion. In the event of significant differences being detected, the data were subjected to post-hoc analysis using Student-Neumann-Keuls method.

3. Results

The relationship between stride velocity and ankle dorsiflexion range can best be described by an exponential relationship (Fig. 1) with

$$Velocity = 0.6531 \times e^{0.05 \times angle} - 0.196 \ (\mathrm{m\ s^{-1}}) \quad (1)$$

This regression demonstrated a significant correlation ($R^2 = 0.472$; $P < 0.001$). No significant correlation could be found between ankle range and cadence, irrespective of the nature of the regression applied ($R^2 < 0.15$).

Step length on the uninjured side showed a significant correlation with ankle range (Fig. 2) ($R^2 = 0.422$; $P < 0.001$). In this case, the best fit model was a sigmoid relationship, where

$$Step\ length = 625/[1 + e^{(2.68 - angle)/5.09}] \ (\mathrm{mm}) \quad (2)$$

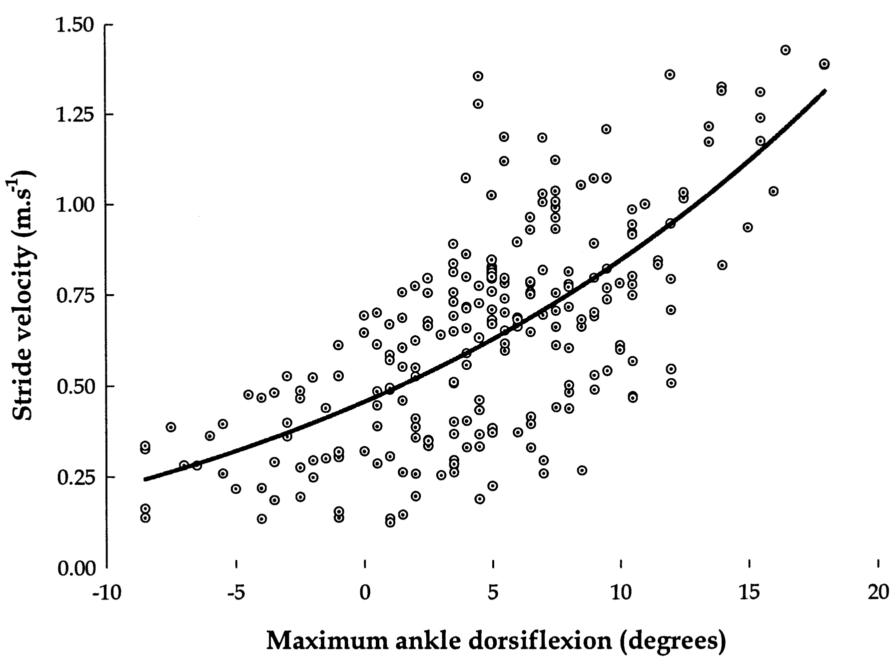

Fig. 1. Effect of maximum ankle dorsiflexion range on stride velocity. Zero degrees represents a plantigrade ankle. All observations ($n = 239$) illustrated. Regression derived from Eq. (1) (see text).

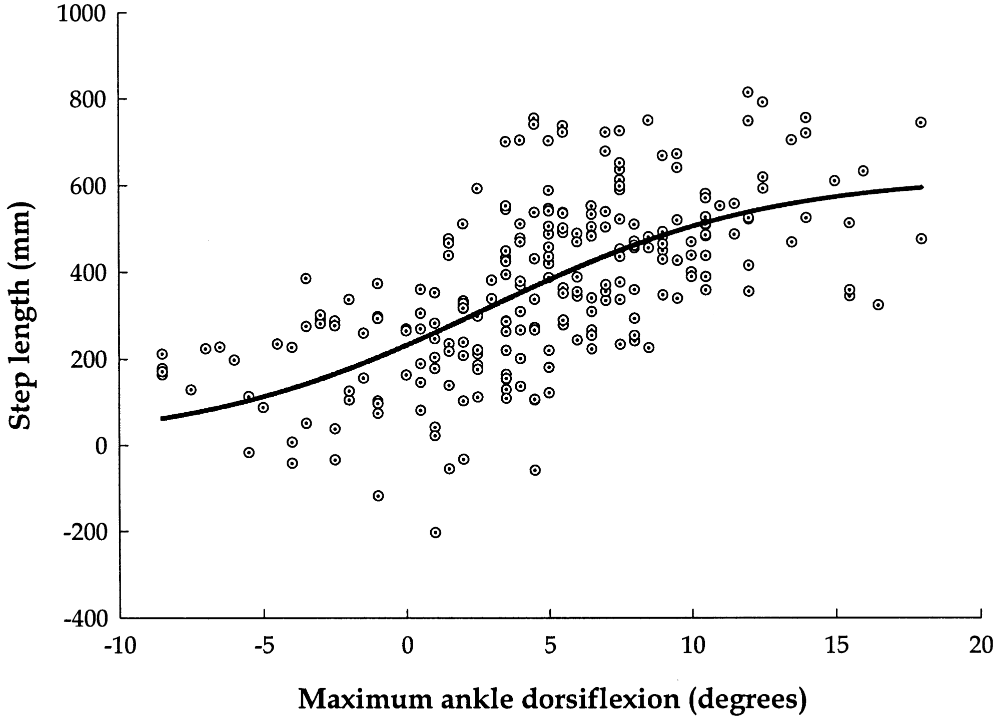

Fig. 2. Effect of maximum ankle dorsiflexion range on uninjured side step length. Regression derived from Eq. (2) (see text).

The appropriateness of this particular model will be discussed below.

Single support time on the injured side was strongly correlated with stride velocity ($R^2 = 0.728$; $P < 0.001$) through a logarithmic regression, with

$$\textit{Single support time} = 44.5(1 - e^{-1.73 \times \textit{velocity}}) \quad \text{(\% cycle time)} \qquad (3)$$

Examining the effect of ankle range on the symmetry of step length (Fig. 3), it was found that a significant difference existed between symmetry values for ankle ranges below 4° of dorsiflexion and those above that value ($P < 0.05$).

The values obtained for single support symmetry demonstrated the same variance patterns as those obtained for step length symmetry, although the magnitude of the ratio was more distinct in the case of step length (Fig. 3). Here, too, groups with ankle range less than 4° of dorsiflexion were significantly less symmetrical than those with greater than 4° available dorsiflexion ($P < 0.05$). The symmetry of double support, however, was not found to vary between groups ($P = 0.643$).

Examination of the general temporal phase relationships indicated that similar patterns were observable with respect to the influence of ankle range on temporal phases as were found with symmetry values (Fig. 4). The double and single support phase proportions, approximating those seen during normal walking, were achieved only when ankle dorsiflexion reached about 8°.

Relationships between stride velocity and the spatial parameters corresponded to a simple linear model in all cases:

$$\textit{Stride length} = 0.874 \times \textit{velocity} + 0.279 \text{ (m)} \quad (R^2 = 0.831;\ P < 0.0001) \qquad (4)$$

$$\textit{Injured side step length} = 0.304 \times \textit{velocity} + 0.277 \text{ (m)} \quad (R^2 = 0.602;\ P < 0.0001) \qquad (5)$$

$$\textit{Uninjured step length} = 0.57 \times \textit{velocity} + 0.004 \text{ (m)} \quad (R^2 = 0.747;\ P < 0.0001) \qquad (6)$$

4. Discussion

It has been shown that around 10° of ankle dorsiflexion are necessary for level walking in able-bodied subjects [15]. Interestingly, this range is not velocity dependent to the same extent as is found with the hip and knee joints. Between walking speeds of 0.8 and 1.9 m s^{-1}, ankle dorsiflexion remains relatively constant [16,17]. The reason for this may be partly because the physiological range of dorsiflexion in healthy adults is limited to 18° (S.D. 7.4) [18], and also because continued dorsiflexion in late stance would locate the centre of pressure in an extreme position anterior to the ankle, creating a large external moment at that joint. This moment would place a consequently greater demand on the calf muscles, which would be required to generate maximum force in a fully stretched position. Thus subjects may preferentially allow the heel to lift in order to reduce the demand on the calf muscles.

The main results of this study suggest that, not unexpectedly, maximum available ankle dorsiflexion range is influential in determining the contralateral step length. The overall regression between ankle angle and step length across the entire data set is best described by a sigmoid curve. The portion of the curve corresponding to the most rapid change in step length lies between 0 and 10°, a region which also marks the transitions from asymmetry to relative symmetry in the temporal and spatial parameters and the approach of the temporal phase values towards those associated with 'normal' gait.

At values less than 0° of dorsiflexion, gains in ankle range are not associated with marked increases in walking speed. Pain is likely to be the predominant constraining influence at this stage, as evidenced by the significant lack of symmetry of single support time for values of ankle dorsiflexion below zero. As shown in Fig. 4, the uninjured single support phase (corresponding to the injured swing phase) stabilises relatively quickly, even before a plantigrade ankle position can be achieved, whereas the injured side single support time shows a progressive improvement from extremely low values across much of the test result.

Subsequent to the attainment of 10°, further gains in ankle range make little impact on step length. The need for more ankle range is, as has been demonstrated in healthy subjects, less important. Walking speed, however, continues to increase at an exponential rate after the step length stabilises, which may be a function of improvements in pain and, perhaps, confidence.

The linear relationships between step lengths and walking speed have differing coefficients for the uninjured and injured limbs. The predicted point of convergence of these regressions (from Eqs. (5) and (6)) would be at a walking speed of 1.03 m s^{-1} and a step length of 0.59 m. From Figs. 1 and 2, it is apparent that these conditions are satisfied when the ankle is able to dorsiflex to approximately 10°. The fact that there are few data points beyond this walking speed, because subjects were discharged from further treatment by this time, means that these linear models are only valid for slow, asymmetrical walking. Beyond the point of convergence of the two limbs the regression would change to a different, probably curvilinear relationship.

On the basis of these data, it would appear that the resolution of ankle sprain, as evidenced by the recovery of pain-free ankle dorsiflexion range of motion, is attended by a consistent response from the temporal and spatial parameters of gait.

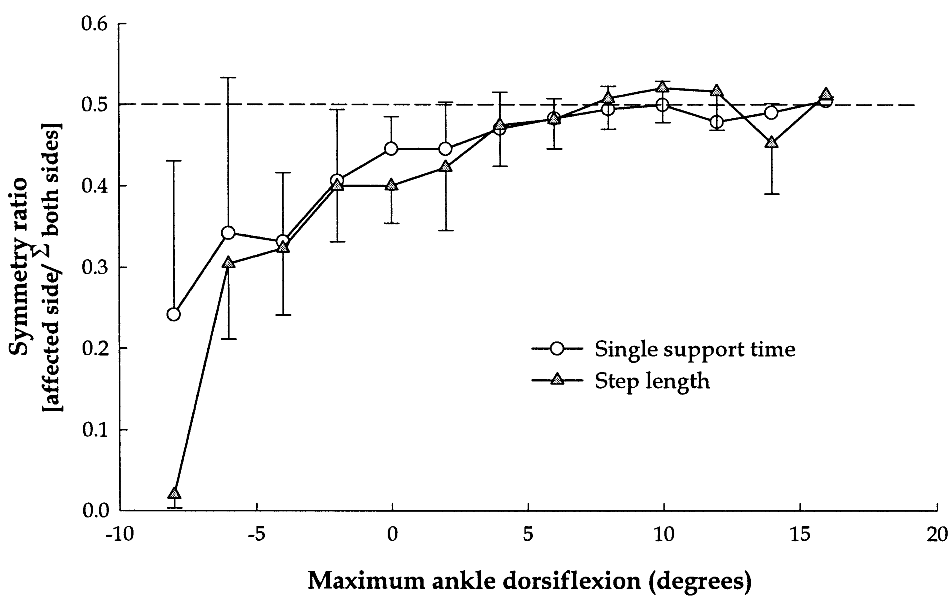

Fig. 3. Effect of maximum ankle dorsiflexion range on symmetry of gait variables. Mean and one standard deviation illustrated. Open circles represent single support time ratio; filled triangles represent step length ratio. Data are grouped according to ankle dorsiflexion range in 2° increments.

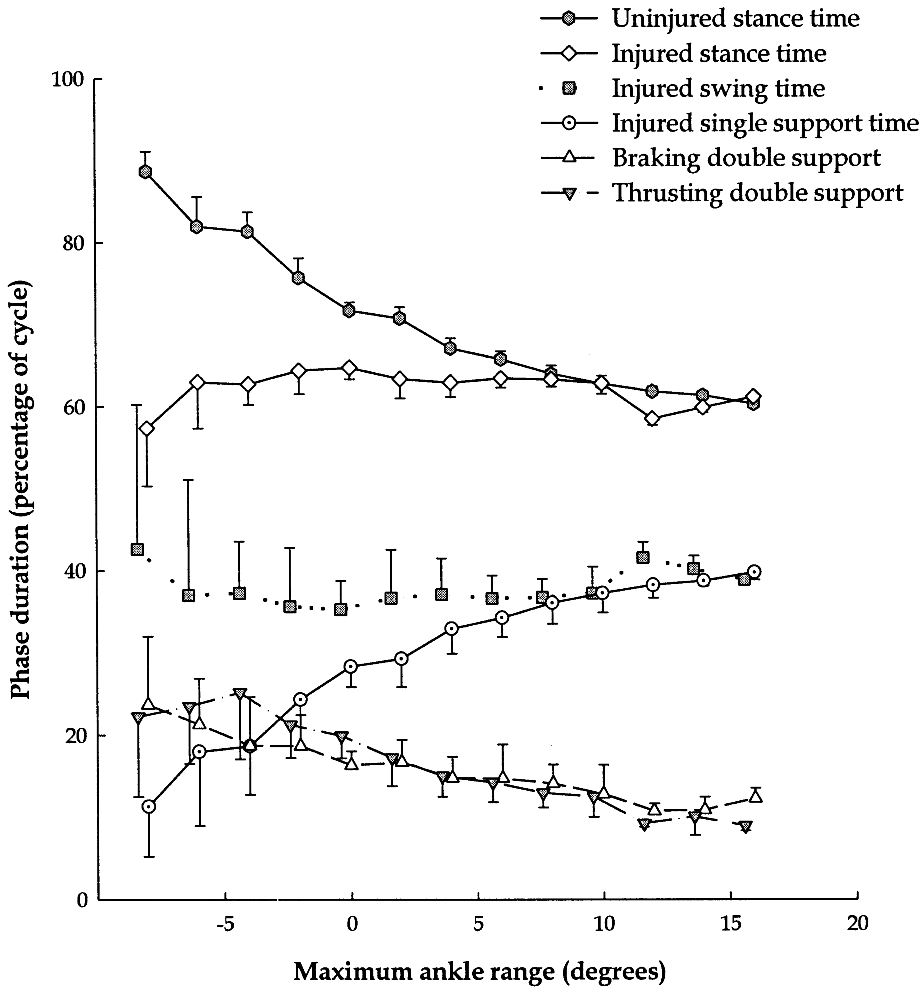

Fig. 4. Effect of maximum ankle dorsiflexion range on temporal relationships of the gait cycle. Mean and standard deviation values illustrated. Grouping of data follows convention described in Fig. 3.

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