Clinical Biomechanics 33 (2016) 73–78

Contents lists available at ScienceDirect

Clinical Biomechanics

journal homepage: www.elsevier.com/locate/clinbiomech

Effects of ankle instability on running gait ankle angles and its variability in young adults

Daniel Hamacher [a,*], Karsten Hollander [b], Astrid Zech [a]

[a] Institute of Sport Science, Friedrich Schiller University of Jena, Seidelstraße 20, 07749 Jena, Germany
[b] Department of Sports and Exercise Medicine, Institute of Human Movement Science, University of Hamburg, Turmweg 2, 20148, Hamburg, Germany

ARTICLE INFO

Article history:
Received 15 September 2015
Accepted 11 February 2016

Keywords:
Running
Proprioception
Kinematics
Ankle injuries
Variability

ABSTRACT

Background: Considering that proprioceptive deficits seem to be responsible for chronic ankle instability, the precise control of ankle angles during running may be impaired. Thus, the aim of the study was to evaluate the influence of chronic ankle instability on intra-individual variability of ankle kinematics during running.
Methods: Lower extremity running gait kinematics of 12 recreational athletes with chronic ankle instability (mean age: 24 years, SD: 3 years; strides analyzed: 40.0, SD = 1.7) and 12 matched healthy controls (mean age: 27 years, SD: 6 years; strides analyzed: 40.2, SD = 2.5) were registered on a treadmill. Mean ankle angles (inversion/eversion, plantarflexion/dorsiflexion) and intra-individual standard deviations (variability) were calculated at each percent of the running gait cycle. Group differences were examined using statistical parameter mapping. To estimate effect sizes, Hedges' g was calculated.
Findings: No group differences in the inversion/eversion or plantar-/dorsiflexion ankle angle were found. The inversion/eversion variability was significantly higher ($P < .050$) in individuals with chronic ankle instability during the stance and swing phase. The highest Hedges' g values were registered at 15% ($g = 0.575$, $P < .000$) and 95% ($g = 0.551$, $P = .002$) of the running gait cycle. The plantar-/dorsiflexion ankle angle variability showed no significant differences.
Interpretation: Patients with chronic ankle instability exhibit a higher variability of ankle kinematics during running. This indicates altered sensorimotor control which is probably an underlying mechanism of chronic ankle instability. Thus, variability measures may help to better quantify treatment effects in future.

1. Introduction

Among the lower extremity injuries recorded by emergency departments in the United States, ankle sprain is the most frequently occurring, particularly in young adults and teenagers (Hertel, 2002; Lambers et al., 2012). Almost half of these injuries occur during sports activities (Waterman et al., 2010). Although most of these patients return to their original sports activities (Anandacoomarasamy and Barnsley, 2005), their symptoms, such as weakness, pain, swelling, laxity, functional limitations, instability, and giving-way persist several months or years post-injury (Anandacoomarasamy and Barnsley, 2005; Konradsen et al., 2002; Verhagen et al., 1995). As a result, chronic ankle instability (CAI) may be developed and about 20% of patients with ankle sprain report the occurrence of recurrent ankle sprains (Hertel, 2002; Konradsen et al., 2002).

It has been sufficiently shown that people with CAI have an impaired proprioception (measured in force and angle reproduction tests) (Docherty and Arnold, 2008; Freeman et al., 1965; Konradsen and Magnusson, 2000) which could result from a damage to muscle mechanoreceptors and from ligamentous injury after ankle sprain (Docherty and Arnold, 2008). Deficits in proprioception may also be responsible for the giving-way sensation (Freeman et al., 1965). Taking additional effects other than proprioceptive deficits in CAI into account (e.g. central adaptations due to lower extremity injuries, Kapreli and Athanasopoulos, 2006), it is not surprising that subjects with CAI depict not only differences in static and dynamic postural control (Hoch et al., 2012; McKeon and Hertel, 2008) but also an adapted gait pattern. Affected people walk slower with lower cadence and step length (Gigi et al., 2015). In particular, CAI patients have a reduced impact in the stance phase and a lateral shifted body weight (Nyska et al., 2003). In 2002, a first cadaver study of simulated gait illustrated that at heel strike, the ankle joint exhibits a high degree of intrinsic stability even in unintentional mal-alignment (Konradsen and Voigt, 2002). The authors further exposed that a collision in late swing phase could lead to a high risk of ankle sprain. Other studies on subjects with CAI reported significant differences in the frontal plane ankle angles, sagittal plane ankle angles (Chinn et al., 2013; Monaghan et al., 2006), and shank rotation as compared to healthy controls (Drewes et al., 2009a), while the hip and knee kinematics remained unchanged (Monaghan et al., 2006). Conflicting results were found regarding changes in ankle inversion/eversion during gait (Chinn et al., 2013; Drewes et al., 2009a; Foss

* Corresponding author.
E-mail address: Daniel.hamacher@uni-jena.de (D. Hamacher).

http://dx.doi.org/10.1016/j.clinbiomech.2016.02.004

et al., 2009; Monaghan et al., 2006; Ridder et al., 2013). Regarding the frontal plane ankle kinematics, decreased variability (as measured by sample entropy) was found in subjects with CAI. This phenomenon was discussed to be the consequence of a decreased capability to adapt and flexibly react to perturbations (Terada et al., 2015). Nevertheless, nearly all existing studies indicate that CAI is strongly associated with changes in ankle gait kinematics and kinetics.

In running gait, CAI groups were more plantar-flexed in swing and stance phase (Chinn et al., 2013; Drewes et al., 2009b). In the frontal plane, subjects with CAI depict a more inverted ankle angle in stance and swing phase (Chinn et al., 2013; Drewes et al., 2009a; Lin et al., 2011). However, a study using a rigid foot model did not find any differences in the stance phase (Ridder et al., 2013). Remarkably, in most studies, the differences in sagittal and frontal plane were often not found at heel strike or at the maximum ankle angles (Brown et al., 2008; Chinn et al., 2013). Regarding the ankle variability, a vector coding approach did not reveal any differences comparing CAI subjects with healthy controls (Herb et al., 2014). To our current knowledge, there are no further studies investigating changes in running gait ankle angle variability. Considering that proprioceptive deficits seem to be responsible for CAI, it is reasonable that the reproducibility of ankle angle kinematics during running is impaired to a certain extent. In gait analyses, motor control is frequently quantified using variability measures (Hausdorff, 2005). If diminished motor control is an issue in running gait of patients with CAI, higher ankle angle variability should be observable. However, these considerations remain speculative since no relevant data are available in the literature. Thus, the aim of the current study was to identify the magnitude of changes in ankle kinematics during running in CAI patients compared to healthy controls by analyzing not only the mean ankle angles but also the variability of ankle angles during subsequent strides.

2. Methods

2.1. Subjects

Twenty-four recreational athletes (running, team sports) were included. Data of 12 subjects with unilateral functional ankle instability (CAI, 10 females and 2 males, age: 24 [SD: 3] years, weight: 69.5 [SD: 12] kg, affected side: 8× right ankle, 4× left ankle) were matched for gender, age, and weight to healthy individuals (10 females and 2 males, age: 27 [SD: 6] years, weight: 65.8 [SD: 10] kg) using the propensity score matching approach. Inclusion criteria were a regular participation in recreational sports activities and for the CAI group a unilateral functional ankle instability based on criteria (previous ankle sprain, repeated giving way and sprain) proposed by Hubbard and Kaminski (Hubbard and Kaminski, 2002). The subjects of our CAI group had at least one moderate to severe inversion ankle sprain within the last 5 years leading to an absence of sports activities for more than 8 days. In the last 12 months, the CAI subjects self-reported a minimum of two recurrent sprains or the feeling of "giving-way." Exclusion criteria were a bilateral ankle injury and all other lower extremity injuries in the last 12 months. Except for functional ankle instability in the corresponding group, no further acute motor-functional impairments or medication that could influence running gait kinematics have been self-reported by the subjects. The study has been approved by the ethics committee of the medical association Hamburg (protocol no. PV4271) and followed the principles of the Helsinki Declaration. Furthermore, all subjects provided written informed consent to their voluntary participation in this study.

2.2. Testing procedure

The calibration procedure consisted of a dynamic and a static calibration using a calibration kit (VICON, Oxford, UK). Afterwards, standing calibrations were performed prior to the testing trial for each participant to create a biomechanical lower body model. After an accommodation and warm-up period at self-selected walking speed on a treadmill (Ergo-Fit TRAC 4000, ERGO-FIT GmbH & Co. KG, Pirmasens, Germany), three-dimensional kinematic data were registered with an 8-camera infrared motion analysis system (VICON, Oxford, UK) at a sampling rate of 200 Hz. According to the Plug-in-Gait model (VICON, Oxford, UK) which have been successfully applied in prior running gait studies in healthy subjects (Hollander et al., 2015; Hollander et al., 2014) and subjects with CAI (Chinn et al., 2013), sixteen retro-reflective markers (14 mm diameter) were attached onto specific bony landmarks. All participants ran at 2.78 m/s for a 60 s period. Within this period, the second 30 s were recorded. While running, all subjects wore an Asics GT-2160® (ASICS, Kobe, Japan) to standardize the footwear. This cushioning running shoe is characterized by an ethylene–vinyl acetate midsole, medial arch support, 12 mm heel-forefoot offset and weighs 314 g (women's US size 6.5).

2.3. Data analysis

Data acquisition and processing were performed with Vicon Nexus software (version 1.7.1 VICON, Oxford, UK). After filtering the kinematic data by means of the Woltring filtering routine (mean square error = 15), the inversion/eversion as well as the plantar-/dorsiflexion ankle angle as a function of time were exported to csv files. Further processing was done with MATLAB software (version R2014a, MathWorks, Natick, MA, USA). Foot strikes were defined at the time points when the sign of vertical velocity of the distal heel marker changed from negative to positive (Fellin et al., 2010). Each running gait cycle was time normalized (natural cubic spline interpolation) to 101 points with the first and last point reflecting a heel strike. Thereafter, mean angles and its intra-individual stride-to-stride standard deviation (as measures of running gait variability) of each percent of running gait cycle were calculated for the affected and unaffected ankle.

2.4. Statistics

To assess significant between-group differences, the statistical parametric mapping method was applied using an open-source software package (version 0.3) (Pataky, 2012). Initially, the test statistics for each time normalized data point was determined by means of Student's independent samples t-test. To avoid the problem of multiple testing, statistical parameter mapping was deployed. This approach uses the random field theory to account for spatiotemporal correlations of kinematic data. As a result, the significances of threshold clusters are calculated (Pataky, 2012). In addition to quantify effect sizes, Hedges' g was calculated.

3. Results

For CAI subjects, we included 40.2 (SD: 2.5) and for the healthy controls 40.0 (SD: 1.7) strides into the analysis. Our results do not expose differences between CAI patients and healthy controls in the inversion/eversion or plantar-/dorsiflexion angles for both the effected and unaffected ankle at any time point of the running gait cycle (Fig. 1). Intra-individual inversion/eversion variability over all strides (Fig. 2) was significantly higher in CAI patients in stance phase (11–24% of the gait cycle, $P < .000$) and swing phase (77–83% of the gait cycle, $P = .005$; 92–97% of the gait cycle, $P = .007$) of the affected ankle. Regarding the unaffected ankle, only a higher frontal plane ankle angle variability was observed in swing phase (66–69% of the gait cycle, $P = .023$). The highest Hedges' g effect sizes were registered in stance phase at 15% ($g = 0.575$) and swing phase at 95% ($g = 0.551$) of the running gait cycle of the affected ankle. The plantar-/dorsiflexion ankle angle variability did not show any statistically significant differences.