Peroneal reaction time measurement in unipodal stance for two different destabilization axes

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Background: The variability of peroneal reaction time measurements is a major problem when using this parameter to control rehabilitation or proprioceptive training processes. In order to control peroneal reaction time values, some extrinsic factors should be considered. The purpose of this study was to measure peroneal reaction time in unipodal stance for two different destabilization axes.

Methods: The peroneal reaction time of 10 healthy subjects was measured from kinematic and electromyographic data in an experimental study using an ankle destabilization device.

Findings: In a preliminary analysis, results showed that the destabilization axis orientation did not affect peroneal reaction time values (68.5 ms, standard deviation = 9.5 ms and 71.5 ms, standard deviation = 8 ms) for destabilizations in the frontal plane and around the Henke’s axis, respectively). However, the inter-trial variance of inversion velocity peaks explained between 40% and 49% of the peroneal reaction time variance. When trials were selected on the basis of homogeneous inversion velocity peaks, results showed that peroneal reaction time values for the peroneus brevis were shorter during inversion movements performed around the physiological Henke’s tilting axis (63 ms, standard deviation = 9 ms vs. 71 ms, standard deviation = 8 ms).

Interpretation: Our findings evidenced that tilting axis orientation must be considered as an extrinsic factor that may influence peroneal reaction time. Moreover it also seems necessary to consider inversion speed values to adequately compare peroneal reaction time values.

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1. Introduction

The possibility of a link between longer peroneal reaction times (PRT) and an increased risk of lateral ankle sprain was postulated by Wilkerson and Nitz (1994). Since, many studies demonstrated that functional instability of the ankle was associated with an increase of PRT (see Menacho et al., 2010 for a review). However the inter-trials PRT variability appears as a major problem to compare PRT values between conditions or groups. As underlined by Caufield cited by Delahunt (2007) “effort should be made to standardize subject classification in studies on ankle instability, as unless this is done it will be very difficult to interpret research findings” (pp100 lines 52–56). In other words, if standardized measurement conditions were provided, the peroneal reaction time can be effectively used as a clinical parameter in rehabilitation and/or proprioceptive training programs. Previous studies have shown that PRT is a parameter independent of anthropometric factors, positively influenced by plantarflexion and negatively influenced by neuromuscular fatigue (Benesch et al., 2000). However, in order to control PRT values, additional extrinsic factors should also be considered.

Following Hertel (2002), ankle sprain motion is not a pure single-plane motion but it is associated with motions in the two other planes. According to the orthopaedic definition, lateral ankle sprain motion is a combined plantarflexion (sagittal plane), adduction (transverse or coronal plane) and supination (frontal plane). This particular movement, named inversion, revolved about the functional axis of the subtalar joint also called Henke’s axis. However, different devices as trap-door or tilting inversion platforms inducing ankle motions restricted to the frontal plane were currently used to measure PRT values. In order to reproduce more realistic inversion trauma mechanisms, some authors used modified inversion platforms (Eechaute et al., 2007; Mitchell et al., 2008; Vaes et al., 2002). Unfortunately, in such conditions the feet were in an initial position different from the neutral–anatomical position leading to PRT modifications (Benesch et al., 2000). Recently Chan et al. (2008) described a mechanical sprain simulator. The destabilization axis could be adjusted in order to gradually provide pure supination or plantarflexion motions. Such material allows kinematic studies of a wider variety of sub-injury ankle sprain motions. Finally, because initial foot position and ankle movements were different depending on the nature of the device, it could be hypothesized that cutaneous, capsular and muscular structures acting around the ankle were differently stretched during the destabilization movement. This may lead to different afferences generating different peroneal reaction times. In keeping with the neuronal population vector model (Georgopoulos et al., 1986), Bergenheim et al. (2000) have shown
that movement direction is encoded by population of muscle spindle afferents originating from all the muscles surrounding a particular joint and subject to deformation. In addition, Aimonetti et al. (2010) showed that some cutaneous afferents are also involved in directional movement encoding. As a result, it could be hypothesized that, because cutaneous, capsular and muscular materials acting around the ankle joint will be optimally stretched, an ecological (Henke) destabilization axis will be associated with shorter peroneal reaction times. In other words, PRT value discrepancies could be explained in part by the ankle destabilization axis.

The purpose of the present study was to add original data for peroneal latencies measured in unipodal stance for two different destabilization axes.

2. Methods

2.1. Subjects

A group of 10 healthy active subjects (6 men and 4 women; mean age: 24.8 ± 6.4 years, mean height: 177 ± 7.3 cm, and mean weight: 73 ± 8.4 kg) with no ankle instability and no known history of neurological or motor dysfunctions participated in the study. The subjects' informed consent was obtained. The study was approved by the local research ethic committee of the University of Savoie in accordance with document number 2010-LPE-E3NF.

2.2. Task and apparatus

This experiment consisted to induce ankle destabilizations to assess peroneal reaction time (PRT) values. Subtalar joint destabilizations were induced using a shoe equipped with a mechanical articulator mounted under the heel. We used a custom version of the device previously described by Forestier and Toschi (2005) and illustrated in Fig. 1. An electric switch controlled an electromagnet that triggered the release of the articular device. Subjects were asked to stand on their dominant (right) leg with full knee extension. The left foot was lifted so that subject’s big toe touched the medial malleolus of the supporting leg. The dominant leg was operationally defined as the leg with which the subject would kick a ball. The right leg was dominant for all the subjects. This position allowed us to control that 100% of body weight was applied on the right ankle when destabilization occurred.

Fig. 1. Illustration of the device used to induce ankle destabilizations. Upper panel: during one of the two experimental conditions, destabilizations of the subtalar joint were induced around the Henke’s axis, with an inclination of 42° in the frontal plane. The left picture illustrates the ankle position before destabilization whereas the right picture illustrates the ankle position after destabilization. Lower panel: during the other experimental condition, destabilizations of the subtalar joint were induced in the pure frontal plane. The left picture illustrates the ankle position before destabilization whereas the right picture illustrates the ankle position after destabilization.
The potential influence of destabilization axis orientation on peroneal reaction time values was assessed through two different experimental conditions. Ankle destabilizations were induced (i) in the frontal plane and (ii) around the Henke’s axis, i.e. with an inclination of 42° in the frontal plane. While frontal destabilizations have been mostly used to assess peroneal reaction times, destabilizations around the Henke’s axis allowed mimicking ecological inversion movements of the subtalar joint. The ankle displacements were recorded by means of a potentiometer mounted on the articulator axis. The activities of the peroneus brevis (PB) and peroneus longus (PL) muscles were recorded by surface electromyography (EMG). The skin was shaved and cleaned with an alcohol–ether–acetone solution to minimize impedance. According to the recommendations of Cram et al. (1998), surface electrodes (Skinact™ Ag-AgCl electrode type F55) were placed with a 2-cm inter-electrode distance longitudinally over the bellies of the right PB and PL. The EMG signals were preamplified (×375) close to the recording site; band-pass filtered (8–500 Hz), and amplified (×412) in a special measurement unit (Mega™ Electronics, Finland).

Kinematic and electromyographic signals were transferred to a 12-bit A/D conversion acquisition card (Measurement Computing™ Model PCI-6052E) in order to be synchronized and sampled at 1000 Hz (Dcoll™, GRAME, Quebec). These synchronized signals were then recorded for subsequent analyses using custom software developed in Matlab™ (Analyse™, GRAME, Quebec).

2.3. Procedure

Each subject began a testing session with a slight warm up consisting of alternative inversion/eversion movements. A typical experimental session for a single subject consisted of a total of 20 ankle destabilization trials: 10 in the frontal plane and 10 around the Henke’s axis. The orientation of the articulator device mounted under the heel was modified between experimental conditions. The order of presentation of experimental conditions was randomized between subjects so that half received one order and half received the other order. As stipulated above, subjects were asked to stand on their right leg to control that 100% of the body weight was applied on the destabilized ankle. Moreover, this position was chosen because most of lateral ankle sprains occur during the single support phase of gait cycle. A light touch of the right forearm on an adjustable support (shoulder height) was allowed to facilitate subjects’ equilibrium. In this position, they were instructed to relax and close their eyes in order to minimize the influence of visual cues. They also wore noise-cancelling headphones in order to minimize the influence of auditory cues. EMGs signals were displayed on-line to the experimenter by means of an oscilloscope in order to control the EMG background noise. When stabilized, the experimenter induced the ankle inversion. Subjects were then allowed to put their left foot on the floor and were instructed to “react naturally against the ankle destabilization.” To prevent muscular fatigue development, a short rest was allowed between each trial.

2.4. Data analysis

The ankle inversion displacements were filtered (fourth-order Butterworth with a 7 Hz low pass cut-off frequency with dual-pass to remove phase shift) prior to calculation of the inversion velocity (finite-difference algorithm). The beginning of each inversion displacement was automatically determined when it exceeds 1-fold standard deviation of background noise. The inversion velocity peaks were also automatically determined. EMG raw data were full wave rectified and then averaged (25 ms moving average window) to obtain EMG data. Muscular activity onset was automatically detected when muscular activity exceeds 10-fold standard deviation of background noise during the last second before the mechanical destabilization. A visual inspection was also performed in order to prevent any artefacts. As illustrated in Fig. 2, the PRT was defined as the time between ankle inversion and peroneal muscles activity onsets.

Fig. 2. Illustration of the methodological approach used to compute PRT. Upper panel: illustration of a typical ankle inversion movement. The onset time (t0) was defined from inversion displacement signal and EMG signals were synchronized (longer vertical line). Middle and lower panels: Onsets of Peroneus Brevis and Peroneus Longus activities were automatically marked (shorter vertical line). PRT was defined as the delay between ankle inversion and peroneal muscles activity onsets.
2.5. Statistical analyses

The Shapiro–Wilk test was used to check that all dependent variables were normally distributed. PRT values were submitted to a 2 muscles (PB and PL) × 2 destabilization axes (frontal plane and around the Henke’s axis) ANOVA. Then, a forward stepwise multiple regression analysis was used to determine the level of collinearity between inversion velocity peaks and PRT values. Finally, trials were normalized from real traumatic inversion velocity peaks (between 600 and 650° s⁻¹) and selected data were submitted to a 2 muscles (PB and PL) × 2 destabilization axes (frontal plane and around the Henke’s axis) ANOVA. For the ANOVAs, repeated measures on both factors were systematically executed and post hoc analyses (orthogonal planned comparisons) were performed whenever necessary. Data are reported as means and standard deviations and a .05 alpha threshold was adopted throughout.

3. Results

3.1. Effect of the destabilization axis orientation on PRT values

As illustrated in Fig. 3, the ANOVA showed that the main effect of the destabilization axis orientation on peroneal reaction time values was not significant (\(F(1,9)=2.83, p=0.1271\)). Mean PRT values represent 68.5 ms (standard deviation= 9.5 ms) for destabilizations in the frontal plane and 71.5 ms (standard deviation= 8 ms) for destabilizations around the Henke’s axis. The main effect of muscle was significant (\(F(1,9)=5.15, p=0.0494\)) with slightly longer reaction time values for the peroneus brevis than for the peroneus longus (72.5 ms, standard deviation= 7 ms vs. 67.5 ms, standard deviation= 9.5 ms, respectively). The interaction effect of muscles × destabilization axes was not significant (\(F(1,9)=2.5, p=0.1486\)).

3.2. Correlation analyses between peroneal reaction times and inversion velocity peaks

Our results revealed an important inter-trial variance of inversion velocity peaks (597° s⁻¹, standard deviation = 148 ° s⁻¹ and 515° s⁻¹, standard deviation = 143 ° s⁻¹ for destabilizations in the frontal plane and around Henke’s axis, respectively). In order to assess if inversion velocity values could explain a significant part of PRT variance, multiple regression analyses were performed. The Variance Inflation Factor (VIF = (1/(1 – R²))) values were calculated to check for multicollinearity problems. Based on the recommendations from the literature (Neter et al., 1985), any model with a VIF greater than 10 has a problem of multicollinearity. Because VIFs are estimated at 2.032 and 2.36 no multicollinearity was found with our data. Results evidenced significant negative correlations between inversion velocity peaks and PRT values for ankle destabilizations (i) in the frontal plane (\(F(1,98)=80.40, p=0.0001\) and \(F(1,98)=68.73, p=0.0001\) for PB and PL muscles, respectively) and (ii) around the Henke’s axis (\(F(1,98)=93.88, p=0.0001\) and \(F(1,98)=77.57, p=0.0001\) for PB and PL muscles, respectively). As illustrated in Fig. 4, \(R²\) values were included between 0.40 and 0.49, meaning that inversion velocity peaks explain between 40% and 49% of the PRT variance.

Based on biomechanical data relative to a real accidental lateral ankle sprain (Fong et al., 2010) two phases i.e. a risk-developing phase and an injury phase was determined during sprain injury. Interestingly a maximal ankle inversion velocity of 623°/s was reported for the injury case. Our data showed that despite the large inter-trial variability of inversion speeds, the majority of ankle inversion velocity peaks were dispatched between 600°/s and 650°/s. We decided to select these trials for all the subjects in order to normalize inversion velocity, which is a strong factor of PRT inter-trial variability, and to study destabilizations similar to real traumatic inversions in velocity terms. Once again, PRT values were submitted to a 2 muscles (PB and PL) × 2 destabilization axes (frontal plane and around the Henke’s axis).

As illustrated in Fig. 5, the ANOVA showed that the main effect of muscle was significant (\(F(1,19)=7.97, p=0.0109\)) with longer reaction time values for the peroneus brevis than for the peroneus longus (67 ms, standard deviation = 8 ms vs. 61 ms, standard deviation = 12 ms, respectively). The interaction effect of muscles × destabilization axes was also significant (\(F(1,19)=5.3, p=0.0328\)). Post hoc analysis (planned comparison) yielded that reactions time values for the peroneus brevis were significantly shorter during inversion movements performed around Henke’s axis (71 ms, standard deviation = 8 ms vs. 63 ms, standard deviation = 9 ms, respectively) while there is no differences for reaction time values of the peroneus longus.

4. Discussion

The purpose of the present study was to add original data for peroneal latencies measured in unipodal stance for two different destabilization axes. In line with Benesch et al.’s results, our data showed longer reaction time values for the peroneus brevis than for the peroneus longus. According to their interpretation we think that this difference was due to the longer way the effences take to activate the more distal peroneal brevis muscle.

Contrary to our preliminary hypothesis, peroneal reaction times seem to not be dependent to the destabilization axis orientation. According to Lynch et al. (1996) body weight repartition and speed of inversion were closely linked. In most studies dealing with peroneal reaction time measurements, body weight repartition between each plantar support was inaccurately controlled only by means of verbal instructions. For example, some authors instructed their subjects to “stay barefoot on the plate form with feet shoulder width apart in a relaxed stance and body weight spread evenly between both feet” (Mitchell et al., 2008 pp 1516–1517), to “distribute weight evenly between both feet” (Wilson and Madigan, 2007 pp161), or to “stand on the plate form with his/her weight evenly distributed on both legs” (Elbig et al., 1997; Konradsen and Ravn, 1991; Konradsen et al., 1997 cited by Cordova 2010 pp 350). Because the distribution of body weight on the plate form may directly influence PRT values we used full weight bearing on the measured leg. This procedure is also used by others (Eechaute et al., 2007; Vaes et al., 2002) and allows (i) to control that 100% of the body weight was applied on the measured ankle and (ii) to stimulate a more realistic inversion trauma occurring...
during the single support phase of gait. In spite of this precaution, our results showed that the variance of inversion velocity peaks explain between 40% and 49% of the peroneal reaction time variance. In other words, even when the percentage of body weight applied on the destabilized leg is precisely controlled, the inversion velocity peak could hugely vary and, as a consequence, influence PRT value. It is well known that unipodal standing is associated with large postural oscillations (Burdet and Rougier, 2007). That is the reason why, in this particular posture, the release of the articulator could occur when subject's CG projection was in different locations (e.g. in front, back and/or lateral to the center of the shoe) resulting in important inter-trials variations of inversion velocity peaks. Finally, our results suggest that the variability of the inversion velocity peaks should be considered as an additional extrinsic factor which influenced peroneal reaction time values.

When destabilization trials were selected on the basis of normalized ankle inversion velocities ranging from 600°/s to 650°/s, results show a main effect of the destabilization axis orientation on muscular reaction time only for the peroneus brevis. This is in line with Benesch et al.'s results showing that peroneal reaction times decreased when the ankle was tested in 15° plantar flexion. The authors reported that the reduction of reaction time may be due to a pretension of the talo-fibular ligament. Contrary to Benesch et al.'s results our data do not show such reduction for both peroneal muscles. However when analysing Benesh et al.'s data, we could notice that the main reaction time difference concerned the peroneus brevis muscle (66 to 62 ms and 63 to 61 ms for the peroneus brevis and longus, respectively). Finally, our data associated with those of Benesch et al. (2000) suggest that, during sudden ankle inversion destabilizations, peroneus brevis reaction time values are more influenced by extrinsic factors than peroneus longus ones. Such results seem to support the hypothesis, initially based on anatomical observations, that the peroneus brevis has a key role to lock the subtalar joint.

Two non-exclusive and maybe complementary ways should be considered to explain PRT reduction. First, the triplanar motion induced by the articulator should certainly stretch the joint capsule as well as the anterior talofibular ligament (ATFL) preventing excessive inversion and

![Fig. 4. Relationship between inversion velocity peaks and PRT values for destabilizations (i) in the frontal plane (A and B for the peroneus brevis and longus, respectively) and (ii) around the Henke's axis (C and D for the peroneus brevis and longus, respectively).](image)

![Fig. 5. Peroneal reaction time values as a function of (i) muscles and (ii) destabilization axis orientation. Only trials with inversion velocity peaks between 600°/s and 650°/s were considered for this analyze (see the text for more details). In black, destabilizations in the frontal plane (Peroneus Brevis: 71.5 ms, standard deviation: 8.5 ms and Peroneus Longus: 61 ms, standard deviation: 11 ms). In grey, destabilizations around the Henke's axis (Peroneus Brevis: 63 ms, standard deviation: 9 ms and Peroneus Longus: 62 ms, standard deviation: 13 ms). Note that peroneus longus values were significantly shorter than peroneus longus ones and that peroneus brevis values were significantly shorter for destabilizations around the Henke's axis. (* = P<.05).](image)
internal rotation of the talus on the tibia. As a result, capsular and
tendinous proprioceptive afferences increasing may be associated with
PRT reduction. Second, following Aimonetti et al. (2010), cutaneous
afferents may facilitate the central co-processing of the feedback
information subserving proprioception. The use of a natural destabil-
ization (inversion) movement may be associated with an optimal stretch
of skin region contributing to kinesthesia. Finally it could be hypo-
thesized that a physiological (Henke) destabilization axis leads to reduce
the latency of the muscular response to the mechanical destabilization
because of the optimization of the sensorimotor loop. Therefore, our
data suggest that peroneal reaction time values could be used to assess
the sensorimotor loop efficiency. As suggested by Menacho et al. (2010),
such a parameter could be of great interest for therapists in order to
control ankle rehabilitation programs. However, it is important to note
that some extrinsic factors including the destabilization axis’ orientation
have to be controlled in order to minimize the variability of peroneal
reaction time values. Moreover, the present results suggest that
considering destabilization trials with homogeneous and real traumatic
inversion velocity peaks (between 600 and 650° s⁻¹) is necessary to
calculate peroneal reaction time values between groups or conditions.
Under such conditions, it will be of primary interest to assess if peroneal
reaction times could be used (i) to discriminate healthy subjects from
patients suffering from chronic ankle instability and (ii) to control the
efficiency of ankle rehabilitation programs on recovery of the
sensorimotor loop efficiency.

5. Conclusion
Some studies have proposed that peroneal reaction time values
could be used to control ankle rehabilitation programs (see Menacho
et al., 2010 for a review). However, the variability of such measure-
ments appears as a major problem. In order to minimize peroneal
reaction times variability it is important to control extrinsic factors
such as body weight repartition, initial ankle position or local
muscular fatigue. The present study evidences an additional extrinsic
factor influencing peroneal reaction time values: the destabilization
axis orientation. Furthermore, while extrinsic factors are controlled,
peroneal reaction time variability remains important because of inter-
trial inversion velocity peaks’ variability. Hence, homogenization of
trials from real traumatic inversion velocity peaks seems to be
necessary to compare peroneal reaction time values between groups
or conditions. Finally, it could be recommended for future researches
to systematically control the abovementioned extrinsic factors and to
homogenize the trials from inversion real traumatic velocity peaks.

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