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# Ankle musculature latency measurement to varing angles of sudden external oblique perturbation in normal functionally unstable ankles

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## Abstract

**Background:** Several studies have examined the reflex response of ankle musculature to sudden inversion in noninjured and injured subjects. To date, there have been no studies to determine the effect of versatile degrees & conditions of perturbation on the ankle musculature latency. The purpose of this study was to measure and determine whether there was a difference in ankle musculature latency measurements at 10°, 20°, and 30° in the oblique plane on a dual tilting platform (APS) between normal and functionally unstable ankles under different conditions of perturbation.

**Methods:** The musculature latency of 15 healthy subjects (8 females, 7 males; age range, 18 to 30 years) and 15 patients with functional ankle instability (FAI) (8 females, 7 males; age range, 18 to 30 years) were examined with surface EMG after sudden inversion of the ankle by APS.

**Results**: In all angles of the oblique plane, the latency of Peroneus longus, Tibialis anterior, Peroneus brevis, and Soleus were significantly longer in subjects with unstable ankles under expected and unexpected conditions. Unexpected conditions led to increase the latency of ankle musculature, both for normal and functionally unstable ankles.

**Conclusion:** The significantly longer onset and peak latency of ankle musculature during sudden inversion in the standing position in subjects with unstable ankles is explained by proprioceptive deficit in sensorimotor control of functionally unstable ankles. Unexpected external perturbations of body equilibrium elicit compensatory postural reflexes which cause longer latency of ankle musculature during varying angles of perturbation.

**Keywords:** Latency, Oblique Perturbation, Functional Ankle Instability, EMG

## Introduction

Lateral ankle sprain is one of the most common injuries experienced in sport activities. The recurrence rate for this injury among athletes has been reported to be as high as 80% [1,2]. Recurrent ankle injuries can lead to altered levels of activity and to the individual reporting a functional instability of the ankle. Functional ankle insta-

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bility (FAI) resulting from neuromuscular and proprioceptive deficits is hypothesized to be a major contributing factor to chronic ankle instability [3,4].

Functional instability comprises the patient's subjective feeling of repeated giving way based on proprioceptive disorders or muscle weakness [5]. However, it is often difficult to distinguish between these two entities because a mechanical instability may arise from functional instability [6]. The peroneal longus muscle everts the ankle and is thought to prevent an abnormal degree of inversion during placement of the foot and during adaptation to uneven terrain.

Recently, the measurement of peroneal reaction time or latency to sudden inversion of the ankle on instrumented custom-made platforms (trapdoor) has been frequently advocated to characterize the proprioceptive deficit in FAI [7,10]. Many authors found prolonged peroneal latency in patients with FAI [11,12]. In addition, different experimental methods have been used to measure peroneal reaction time [13,14]. Latency measures in oblique (subtalar) plane as in actual cases of ankle sprain seems to be an important measure for better comparison between FAI and healthy subjects and is not documented to the authors' knowledge.

To date, there have been no studies to determine the effect of smaller degrees of oblique plane perturbation on ankle musculature latency. Furthermore, different angles of perturbation (10, 20, and 30 deg.) under different (expected vs. unexpected) conditions were conducted in the study. Therefore, the purpose of the present study was to determine whether there is a difference in ankle musculature latency measurements at 10°, 20°, and 30° between normal and functionally unstable ankles on a tilting platform - named in this study "Ankle Perturbation System" (APS) to sudden oblique perturbation under variable conditions.

## Methods

*Subjects*: The ankles of thirty (15 healthy, 15 patients with unilateral FAI) young adults volunteers (age = 24.3 ± 4.1 years, wt = 76.0 ± 5.5 kg, ht = 179.5 ± 6.2 cm) were examined (14 male, 16 female). Healthy subjects were excluded if they had a history of lower extremity or spine biomechanical dysfunctions or fractures, neurologic or musculoskeletal disorders, vestibular deficits, obvious LBP within the last 6 months, sensory disturbances of lower limbs, FAI, and ankle injuries within the last 3 months. To be characterized as functionally unstable, the subjects satisfied the following criteria:

(1) experienced at least 1 significant lateral (inversion) ankle sprain of either the right or left ankle, but not both, in which the subject was unable to bear weight or was placed on crutches, within the last year, (2) no reported history of fracture to either ankle, (3) sustained at least 1 repeated injury or the experience of feelings of ankle instability or "giving way" in either the right or left ankle, but not both, (4) not undergoing any formal or informal rehabilitation of the unstable ankle, and (5) have no evidence of mechanical instability as assessed by a orthopedic physician using Anterior Drawer and Talar Tilt Tests. Subjects were pain free and full weight bearing, without a limp, at the time of study. The average time period since their last episode of instability was 6 weeks. The healthy subjects were match paired with the subjects suffering from unilateral FAI. The side of unilateral FAI was matched with the same uninjured side on the healthy subjects. In

addition, height, weight, age, body type, and activity level were used to match the subjects between groups. For the entire group of matched pairs, the average weight difference was 5.1 ± 4.6 kg and the average height difference was 3.4 ± 3.1cm. Subjects were briefed on all testing procedures and asked to read and sign a consent form by a university committee for the protection of human subjects.

*Instrumentation and testing procedure*: Sudden inversion of the ankle was initiated on a tilting platform named Ankle Perturbation System (APS – I.R. Patent 31901) constructed like a trapdoor, which was released by a pneumatic switch through a control panel not observable to the subject. Although the APS was capable to produce frontal, sagittal, and oblique (42° of ankle plantar flexion and 15° of inversion, i.e. “real plane of ankle inversion injury”) perturbations, the data collected from the oblique platform were reported in this study.

All tests were performed in the Biomechanics Laboratory of the University of Social Welfare and Rehabilitation supervised by the Biomechanics Laboratory of the Rehabilitation Faculty of Iran University of Medical Sciences. The tilting angles were adjusted to 10°, 20°, and 30°. The perturbation conditions were repeated both expectedly and unexpectedly. The subjects stood on a custom-designed frontal platform (APS) (Fig. 1a), with both feet tightly fixed on independently movable trapdoors. Each foot was strapped with hook-and-loop tape strips in a foot orthosis, which was fixed to the footplate of the platform in 42° of ankle plantar flexion and 15° of inversion (Fig. 1b). The subject was asked to stand barefoot on the platforms with equal weight on each foot (30 cm apart). The importance of a symmetric posture was also observed.

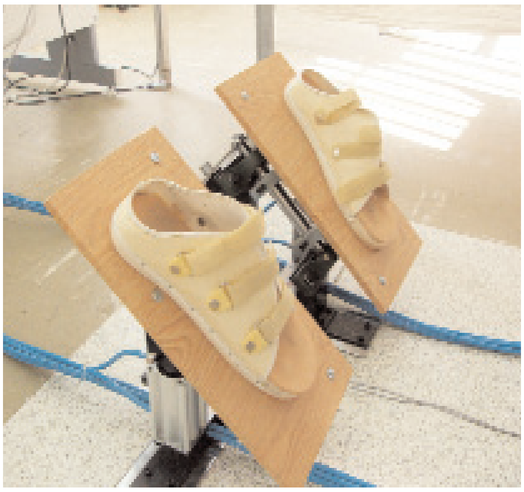

Fig.1a. An Illustration of Oblique Platforms of New Designed Ankle Perturbation System (APS).

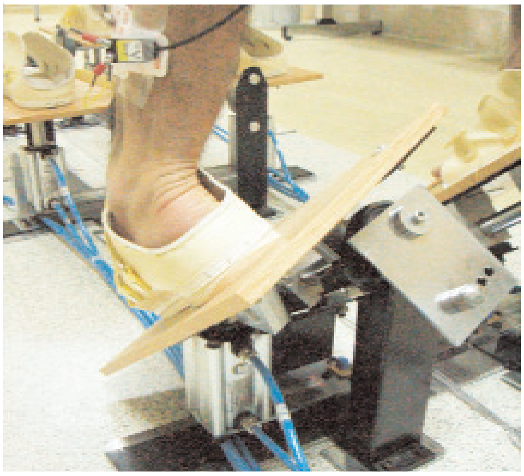

Fig.1b. Perturbation of left ankle of a subject standing on the oblique platform.

The familiarization period included one trial on each leg under unexpected conditions. One of the platforms was released when the scale of APS-Synchronized System software showed equal weight distribution on each platform. All testing was performed with the subject facing a blank screen and looking straight ahead to prevent any visual cuing. Under unexpected perturbation conditions, the subjects did not know which side was to be released. In this situation, the tilting of the platform also occurred without warning, while the subject was blindfolded, wore earphones and listened to music. Earphones have been also utilized to dampen the sound of the platform hitting the end point.

During the perturbation, electromyographic (EMG) activity of the peroneal, tibialis anterior, and soleus muscles were recorded using an electromyograph(CT8 model, MIE Medical Research Ltd., UK) and Ag/AgCl surface bipolar electrodes (13 mm in diameter, 2.5 cm center-to-center distance). The MIE (Measurement Is Evidence) preamplifiers had a gain of 4000, 32 kHz bandwidth, 108 dB (typical) CMRR (Common Mode Rejection Ratio), and 10 [8] ohms input impedance. Electromyographic and load cell [implemented below APS platforms] signals were simultaneously recorded digitally at 1000 Hz as sampling frequency using a 12-bit A/D board and an in-house user-friendly Windows application.

The skin was shaved, degreased with 70% alcohol and prepared with electrolyte gel to minimize resistance. The electrodes were placed over the most prominent part of the muscle bellies of Peroneal, Tibialis anterior, and Soleus muscles. The electrode placements were as follows:

1) Peroneus longus muscle, 3 cm below the head of fibula 2) tibialis anterior muscle, 1cm lateral to the edge of the tibia and 8 cm below the tibial tuberosity 3) Peroneus brevis muscle, 5 cm above the lateral malleolus just behind the fibula 4) Soleus muscle, one-third proximal part of line connecting midway of transverse diameter of popliteal fossa to proximal flare of medial malleolus distal to the lateral head of the gastrocnemius muscle. The same individual performed all electrode placements. Particular movements of the foot verified the correct placement, causing isolated contractions of the muscles to be examined and specific EMG patterns which were observed online on the monitor. To prevent from physiological cross-talk, especially between peroneus longus and tibialis anterior muscles, some strategies such as decrease of inter-electrode distance and contact area of electrodes, and double differential technique are considered and, therefore, a pilot study is performed. Data of the pilot study revealed that there was no cross-talk among above muscles.

*Signal processing, determination of latency, and data management:* At first step, EMG signals were band pass filtered by a 10- to 500-Hz Chebyshev type II filter with no lag (zero phase IIR). The filtration was intended to reduce any motion artifact and high frequency noise contamination of the signals. A 50-millisecond moving window RMS of the EMG signal then was calculated. Because of the nature of RMS process, there was no need to rectify the EMG signals before. The onset of the platform perturbation was first identified from the load cell recording by detecting the time at which there was a significant increase in the force-time slope. Instead of visual detection, an algorithm was used to determine the onset of peroneus muscle activity, which was determined as that signal three standard deviations above the baseline. The muscle response to platform perturbation was considered to occur if the processed EMG signal exceeded 3-fold standard deviation above a baseline mean. A 100-millisecond time window before the platform perturbation was used to calculate the baseline mean. The standard deviation of the 100-millisecond window was used to compute the 3-fold standard deviation threshold. When a response was detected, its latency was measured as the time from the

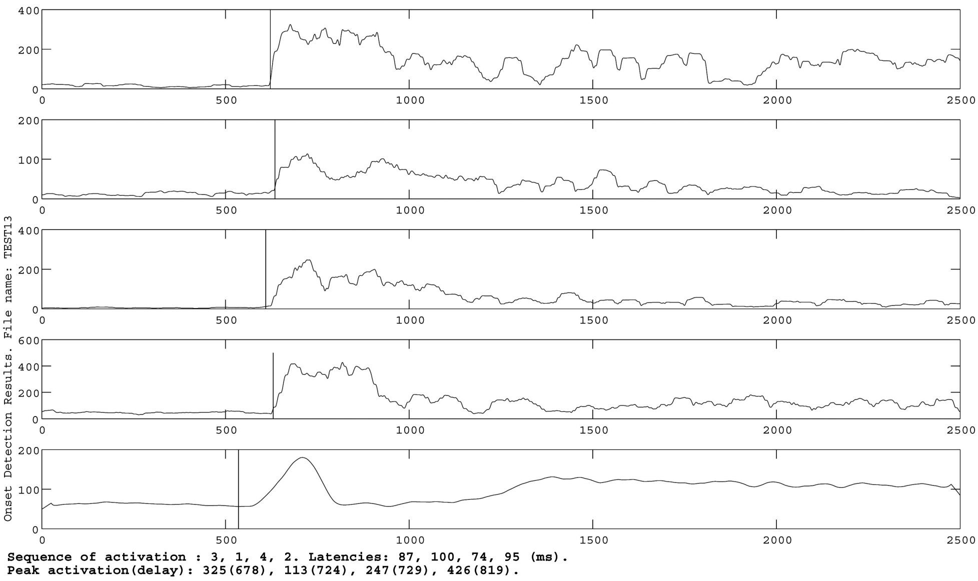


Fig. 2. A Sample of Ankle Perturbation System's Analysis for a trial. Top graph is EMG activity of Peroneus Longus ,the second is for Tibialis Anterior, the third is for Peroneus Brevis, the fourth is for Soleus and the bottom graph is force signal derived from strain-gauge to detect the onset of platform movement. Sequence of activation: 3, 1, 4, 2.Onset Latencies: 87, 100, 74, 95(msec). Peak Latencies: 678, 724, 729, 819 (msec)

start of the platform perturbation to onset of the EMG response (Fig. 2). Data were collected and saved electronically using APS –Synchronization software and transferred to SPSS for Windows 13 (LEAD Technologies, Inc).

*Statistical Analysis:*Differences between the two healthy and FAI groups were tested for significance by means of an independent t test performed for each parameter. The statistical packages SPSS for Windows 13 (LEAD Technologies, Inc) was used for this analysis. The reliability of ankle musculature latency measurement to varying angles of external oblique perturbation was proven by the authors of this article in a methodological study (not published). The ICCs for onset latency of ankle musculature were > 0.96 with the exception of one set of data 30 RE for the Tibialis Anterior (0.92) which may have had some postural factors influence the data. The SEM for all angles and all muscles under both conditions was less than 3.1ms with a mean of 1.1 ms. A statistically significant difference was deemed to occur at an alpha level of 0.05.

## Results

The mean and standard deviation of measured variables (latency) for 24 test positions [3 angles × 2 perturbation conditions (e.g. expected and unexpected) × 2 sides (right and left) × 2 muscles (Peroneal)] are presented in Table 1. This Table also demonstrates mean difference, t statistics, and p values for latency of Peroneus longus and peroneus brevis under different conditions of perturbation. Similar statistical procedures are also performed for tibialis anterior and soleus mus-

| Muscle | Variable | Healthy | | FAI | | Mean Diff. | t | P |
|---|---|---|---|---|---|---|---|---|
| | | Mean | SD | Mean | SD | | | |
| Peroneus Longus | 10RE | 36.8 | 5.21 | 52 | 6.85 | -15.2 | -6.84 | <0.001 |
| | 10LE | 35 | 6.23 | 49.27 | 6.53 | -14.27 | -6.12 | <0.001 |
| | 20RE | 47.27 | 4.27 | 56.93 | 8.63 | -9.67 | -3.89 | <0.001 |
| | 20LE | 45.4 | 5.03 | 56.07 | 8.15 | -10.67 | -4.32 | <0.001 |
| | 30RE | 52.93 | 6.05 | 64.07 | 9.99 | -11.13 | -3.69 | <0.001 |
| | 30LE | 50.07 | 5.26 | 61.8 | 7.95 | -11.73 | -4.77 | <0.001 |
| | 10RU | 45.67 | 4.72 | 63.8 | 9.35 | -18.13 | -6.71 | <0.001 |
| | 10LU | 44.53 | 6.15 | 56.8 | 7.44 | -12.27 | -4.92 | <0.001 |
| | 20RU | 52.87 | 5.59 | 71.87 | 6.93 | -19 | -8.27 | <0.001 |
| | 20LU | 54.67 | 6.9 | 62.87 | 5.77 | -8.2 | -3.53 | <0.001 |
| | 30RU | 61.47 | 5.11 | 79.93 | 8.92 | -18.47 | -6.96 | <0.001 |
| | 30LU | 61.67 | 6.16 | 69.27 | 5.65 | -7.6 | -3.52 | <0.001 |
| Peroneus Brevis | 10RE | 49.53 | 5.68 | 57.13 | 9.49 | -7.6 | -2.66 | 0.01 |
| | 10LE | 48 | 6.29 | 52 | 8.47 | -4 | -1.47 | 0.15 |
| | 20RE | 53.4 | 5.74 | 63.6 | 11.13 | -10.2 | -3.15 | <0.001 |
| | 20LE | 52.6 | 7.31 | 59.33 | 9.66 | -6.73 | -2.15 | 0.04 |
| | 30RE | 58 | 7.15 | 70.73 | 10.48 | -12.73 | -3.89 | <0.001 |
| | 30LE | 58.13 | 7.71 | 64.6 | 8.68 | -6.47 | -2.16 | 0.04 |
| | 10RU | 54.47 | 5.21 | 63.67 | 7.5 | -9.2 | -3.9 | <0.001 |
| | 10LU | 53 | 7.2 | 62.27 | 8.24 | -9.27 | -3.28 | <0.001 |
| | 20RU | 60.8 | 6.72 | 68.07 | 10.67 | -7.27 | -2.23 | 0.03 |
| | 20LU | 59.93 | 6.51 | 68.73 | 9.48 | -8.8 | -2.96 | 0.01 |
| | 30RU | 65.2 | 7.59 | 79.47 | 10.41 | -14.27 | -4.29 | <0.001 |
| | 30LU | 64.73 | 9.03 | 75.20 | 10.62 | -10.47 | -2.91 | 0.01 |

Table 1. Mean, standard deviation, mean difference, t statistics, and P values of ankle musculature for peroneal muscles in healthy and FAI subjects (n=30). Latency of muscular activation following 3 degrees of perturbation (10°, 20° and 30°) in Expected (E) and Unexpected (U) conditions for Right (R) and Left (L) sides have been demonstrated .

cles and shown in Table 2. The FAI group revealed longer latency than the healthy group in most testing situations especially in peroneal and tibialis anterior muscles.

## Conclusion

The study showed longer reaction times of ankle musculature in subjects with unstable ankles. This supports the view of functional instability as a neuromuscular problem so that the term 'neuromuscular instability' appears to describe it more precisely. Before the initiation of the study, the reliability of the ankle musculature latencies in oblique plane was assessed. Under various test conditions, ankle musculature reaction times appear to be highly reliable. Previous reports investigating the peroneus latency to sudden ankle inversion perturbations, mostly in frontal plane, have focused on angles of inversion tilt ranging from 18° to 35°. This large perturbation stimulus results in a high detection rate of muscle activity onset [9,13,15] and may suggest that, at these angles of tilt, the stimulus approaches supramaximal and falls outside the normal range of perturbations at the ankle during activities of daily living and in normal circumstances in sport.

The perturbations used in this study (10 degrees) were similar and are more representative of minor perturbations during functional tasks.

According to smaller angles of perturbation, the results were comparable with the data of Fernandez et al [11]. As in this study, the

| Muscle | Variable | Healthy | | FAI | | Mean Diff. | t | P |
|---|---|---|---|---|---|---|---|---|
| | | Mean | SD | Mean | SD | | | |
| Tibialis Anterior | 10RE | 43.13 | 4.21 | 50.53 | 6.57 | -7.4 | -3.68 | <0.001 |
| | 10LE | 40.33 | 5.46 | 50.33 | 10.38 | -10 | -3.3 | <0.001 |
| | 20RE | 40.07 | 4.18 | 58.53 | 8.48 | -18.47 | -7.56 | <0.001 |
| | 20LE | 39.93 | 5.5 | 56.13 | 10.67 | -16.2 | -5.23 | <0.001 |
| | 30RE | 44.8 | 5.72 | 63.33 | 9.59 | -18.53 | -6.43 | <0.001 |
| | 30LE | 44 | 7.04 | 57.93 | 10.76 | -13.93 | -4.2 | <0.001 |
| | 10RU | 44.33 | 7.65 | 59.93 | 10.12 | -15.6 | -4.76 | <0.001 |
| | 10LU | 45.2 | 6.83 | 50.47 | 9 | -5.27 | -1.81 | 0.08 |
| | 20RU | 47.33 | 4.72 | 65.4 | 11.32 | -18.07 | -5.71 | <0.001 |
| | 20LU | 46.13 | 5.6 | 56.8 | 8.97 | -10.67 | -3.91 | <0.001 |
| | 30RU | 51.67 | 6.94 | 74.73 | 12.6 | -23.07 | -6.21 | <0.001 |
| | 30LU | 51.33 | 8.53 | 62.67 | 8.8 | -11.33 | -3.58 | <0.001 |
| Soleus | 10RE | 58.07 | 4.11 | 68.2 | 4.6 | -10.13 | -6.36 | <0.001 |
| | 10LE | 56.27 | 5.12 | 61.2 | 10.04 | -4.93 | -1.7 | 0.1 |
| | 20RE | 64.47 | 5.94 | 71.33 | 8.49 | -6.87 | -2.57 | 0.02 |
| | 20LE | 62.8 | 6.79 | 65.53 | 9.14 | -2.73 | -0.93 | 0.36 |
| | 30RE | 70.47 | 5.22 | 75.47 | 11.13 | -5 | -1.58 | 0.13 |
| | 30LE | 69.33 | 4.27 | 71.93 | 10.83 | -2.6 | -0.86 | 0.39 |
| | 10RU | 64.8 | 5.29 | 67.87 | 12.59 | -3.07 | -0.87 | 0.4 |
| | 10LU | 63.87 | 6.17 | 70.47 | 5.14 | -6.6 | -3.18 | <0.001 |
| | 20RU | 72 | 5.64 | 74.93 | 13.51 | -2.93 | -0.78 | 0.44 |
| | 20LU | 71.33 | 6.65 | 75.67 | 7.4 | -4.33 | -1.69 | 0.1 |
| | 30RU | 79.6 | 4.93 | 75.73 | 15.09 | 3.87 | 0.94 | 0.35 |
| | 30LU | 78.93 | 5.39 | 82.6 | 7.6 | -3.67 | -1.52 | 0.14 |

Table 2. Mean, standard deviation, mean difference, t statistics, and P values of ankle musculature for tibialis anterior and soleus muscles in healthy and FAI subjects (n=30). latency of muscular activation following 3 degrees of perturbation (10°, 20° and 30°) in Expected (E) and Unexpected (U) conditions for Right (R) and Left (L) sides have been demonstrated.

latency was defined as the difference between the beginning of the tilting movement and the first electrical muscular activity. There is no evidence for comparison of ankle musculature latency measurements between healthy and functionally unstable ankles to different angles of perturbation in oblique plane. Other authors [8, 12] however, found latencies that were far apart from the values obtained in the present study (Table 1). A reason for the discrepancies may be the difference in determination of the onset of muscular activity among researches. For repeatable measurements to be taken accurately, the onsets of the stimulus (platform tilt) and the electromyographic response must be valid. Furthermore, authors, who defined the muscular onset as the point when the EMG activity exceeded a predefined threshold above the background noise, found longer latencies [8,9,15].

This study is the first to show a stimulus threshold response to different ankle oblique perturbations in the ankle musculature latency response. Although this observation is consistent with the previous physiologic interpretations of other muscles crossing the ankle, [16] it remains unclear if it represents a clinically significant difference. In addition, the systematic change of latencies would suggest that the peroneal latency remained within the time frame, which would place it within a polysynaptic (medium loop) reflex window, as described by Lynch et. al [9]. Thus, this study suggests that, in a clinical context, the reflex pathways do not vary greatly, regardless of the angle of tilt.

In addition, it is clear that smaller angles of tilt (10°) are sufficient to trigger a response from the peroneus muscle. The type of sensory receptors that are sensitive to such a small perturbation is open to question. For example, because the relative increase in length of the peroneus muscles is negligible at this angle of tilt, it could be hypothesized that a significant source of the afferent input may be from the joint receptors [17].

Most researchers used trigger activation as the beginning of peroneal latency value recording [12,17]. This does not account for the time difference between signal production at a distal site and initiation of movement of the trapdoor [5]. Although this delay would be in the order of milliseconds, it may be relevant because the time frames being measured are so minute. To eliminate this problem, a strain gauge was used in this study to detect the onset of trapdoor movement.

The study demonstrated the significant difference of peroneal and tibialis anterior latencies between patients suffering functional ankle instability and healthy subjects under different conditions of perturbation. This supports the view of functional instability as a neuromuscular problem so that the term ‘neuromuscular instability’ appears to describe it more precisely. Freeman [3] suggested that functional instability may be the result of impaired co-ordination following articular deafferentation, and that it interferes with the reflexes that depend on articular mechanoreceptors. Vaes et al [8] who were the first pioneers in researching for peroneal reaction times in healthy and unstable ankles especially in oblique plane, used a custom-designed inversion platform which produced sudden 50° of inversion. But their study does not confirm the results of the systematic literature showing a longer latency in subjects with unstable ankles[12,13]. A simple explanation for these controversial reports is that the Vaes et al [18] have used a combination of bilateral and unilateral functional ankle instability in FAI group.

Our protocol was newly developed for measurement of latency of muscular activation in oblique plane and therefore no comparison data were available. Additionally, the versatile test positions were newly considered in this study. One of them was comparison between expected versus unexpected conditions of perturbation. In other words, having or not having information about time and direction of perturbation has been looked in the current study. The results of this study suggest that the latencies of measured variables under unexpected condition were longer than the latencies derived from expected ones in both groups. According to Hassan, [19] the neural responses of the motor control system to mechanical perturbations are designed to ensure stability. Unexpected external perturbations of body equilibrium elicit compensatory postural reflexes causing forward adjustments to overcome the destabilizing effect of expected perturbations.

Finally, in order to control the factors which change the EMG activity characteristics (confounding factors), some optimizations were newly developed such as permanent visual control of weight distribution through scale demonstrated on APS-Synchronization System software, computerized calculation of the latency of muscular activation with respect to tilt signal, the utilization of pneumatic compressor for smoothing the release of platforms and the triple calibration steps which were performed by the examiner before initiation of each trial.

## Conclusion

The current study found significant differences in latency between healthy and functionally unstable ankles and for different amplitudes of ankle inversion perturbation under different conditions. Latency was sensitive to history of injury or subjective inferences of functional instability. Ankle musculature reaction times may help to distinguish between functional ankle instability and normal ankles. Despite these findings, by means of experimental set-up the procedure is time consuming and needs expert knowledge, and more experience is necessary before standard values can be offered.

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