INTRODUCTION
Functional instability at a joint may be defined as impaired proprioception, strength, and postural and neuromuscular control with or without ligamentous laxity [1]. Alternatively functional stability has been defined as possessing adequate stability to perform functional activity resulting from the interaction between the mechanical and dynamic restraints of the joint [2].

Following stroke, hemiparesis is the most common disability, affecting 70% to 80% of stroke survivors [3]. There has been much research into rehabilitation in this area; however there is currently no rigorously designed and evaluated method for measuring shoulder function in the shoulder instability population [4]. This lack of a standardized method has prevented accurate comparisons of treatments [5].

Current measures of rehabilitation following stroke include the Fugl-Meyer assessment of motor recovery, which has been shown to have good validity and reliability [3]. However due to the scoring of each item using only a 3 point scale (0 = can’t perform, 1 = able to perform in part and 2 = able to perform fully) it could be argued this assessment lacks the precision to identify smaller changes.

If, as suggested by Myers [2], functional stability is linked to performing functional tasks, it could be suggested that measuring functional stability of the arm provides a measurement of rehabilitation.

Subluxation is a common measurement of shoulder stability following a stroke. Shoulder subluxation can be measured using X-rays, palpation, arm-length discrepancy and calipers [6]. However, this does not provide a measure of functional stability, as is needed to complete dynamic movements against forces in normal day to day tasks.

One method of quantifying functional stability in the lower limb is the dynamic postural stability index (DPSI), reported by Wikstrom et al. [7], which uses an RMS measure of the variability in ground reaction forces after landing from a jump to quantify instability. This method has been used to compare participants with functional ankle instability and participants with no ankle instability, and found a significant difference between the two groups [7]. One advantage of this technique is that it takes into account stability in all 3 directions, providing more information about possible instabilities in each direction, and then uses this to calculate overall functional stability.

The aim of this study was to quantify differences in arm stability during upper limb postural tasks following shoulder injury. Functional stability was defined similarly to Wikstrom et al. [7], but used displacement from mean position rather than force.

METHODS

Participant’s with and without previous shoulder injury were recruited for a pilot study. Four non-injured shoulders and four previously injured shoulders were measured during the performance of the stability task. A previously injured shoulder was defined as any shoulder that had previous dislocation or subluxation.

The participant, while seated, completed a functional stability task using a Haptic Master robotic arm (Moog FCS, Nieuw Vennep, NL). This functional stability task involved holding the handle of the robotic arm within a target, displayed on a screen. While inside the target area forces were applied through the Haptic Master and the participant was required to resist these forces and minimize the error from the target.

Each participant completed 12 trials in total, 6 with each hand. For each hand, 3 trials were carried out where forces of 5N were applied by the Haptic Master and 3 trials against forces of 10N. During each trial the Haptic Master applied a total of 6 forces, two in each plane of motion (one positive and one negative). The target position was kept constant to control the effects that a different arm position may have on stability. The order of trials was randomized to limit any learning effects.

Throughout the protocol, the force applied by the Haptic Master, the interaction force between the participant’s hand and the Haptic Master the position of the Haptic Master endpoint and the target position were all recorded at a frequency of 50Hz.

EMG data of certain muscles were recorded during the trials at 1000Hz. Correct sites for EMG surface electrodes were found by palpating the subject and cleaned using an alcohol wipe. EMG surface electrodes were placed on: biceps brachii, triceps brachii, deltoid and pectoralis major muscles on the right and left sides of the body.

EMG data were filtered with a highpass Butterworth filter (5Hz) and a low pass Butterworth filter (450Hz) and then rectified. Force data were filtered with a low pass Butterworth filter (20Hz).

The Haptic Master position and force measured data were used to analyze shoulder functional stability in the medial-lateral, anterior-posterior and vertical directions. Stability in the x-direction was defined as (similarly for y and z)

\[
S_x = \sqrt{\frac{1}{N} \sum (\bar{x} - x_i)^2}
\]
Furthermore, to account for the effects of different force levels, compliance of the arm was also calculated using the equation:

$$C_x = \sqrt{\frac{1}{N} \sum \left( \frac{x_i - \bar{x}}{F_i} \right)^2}$$

**RESULTS AND DISCUSSION**

Preliminary data from 1 participant with 2 previously injured shoulders (right numerous subluxations and left a single subluxation) are presented here. The left arm appears to show a lower displacement from the mean than the right arm in the 10N condition (Fig. 1). The lower scores suggest higher functional stability, although we cannot say whether the differences are significant from these pilot data.

The numerous subluxations in the participant’s right shoulder and only a single subluxation in the left shoulder may explain the differences found in functional stability between the two arms. However, as there are currently no data of non-injured shoulders to compare the results to, it is impossible to say if functional stability is lower in this participant’s arms to someone who has had no previous shoulder injuries.

A criticism of the calculation of this stability score is that it does not take into account the changing force levels. A participant could have the same stability score in a 5N and 10N force condition; however in reality it would take more functional stability to maintain the same score at the higher force level.

This is demonstrated with the results for compliance (Table. 1) that take into account the effects of force on the positional displacement. It can be seen that the 10N force conditions have lower compliance than hand-matched 5N conditions. This lower compliance indicates a greater resistance to movement against the force, and suggests greater functional stability.

While it is accepted there are limitations in the approach used, it does provide a simple method for calculating an approximation of arm stability, which could be suitable for a clinical setting. This calculation also provides results in all three planes of movement. Assessing rehabilitation progress in all three planes of motion, as opposed to simply recording the ability to perform or not perform a task assessment, may provide a more accurate measurement of stability, particularly regarding small improvements.

**Fig 1.** Average functional stability results in all planes of movement for left and right hand at 5N and 10N forces. Data were sourced from the entire trial where forces applied by the haptic master were equal to 5N or 10N. Error bars show standard deviation between trials.

**CONCLUSIONS**

This study has proposed a simple way to quantify functional stability. While there are limitations to this method it may be useful in clinical settings, such as quantifying improvements in rehabilitation. Further measurements are ongoing and aim to assess the validity and reliability of this method.

**REFERENCES**


**Table 1.** Average compliance calculated individually for each condition. Lateral, medial, towards, away, upwards and downwards describe the direction of force and the direction of displacement that was used to calculate compliance.