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***In vivo* laboratory validation of the physiometer: a measurement system for long-term recording of posture and movements in the workplace**

Leon Straker^a, Amity Campbell^{a*}, Jemma Coleman^b, Marina Ciccarelli^a and Wim Dankaerts^{c,d}

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Posture and movement are thought to be important risk factors for the development of work-related musculoskeletal disorders. Whole day occupational exposure assessment has typically used self-report or observation techniques, but the need for more accurate measurement is now recognised. The aim of this study was to compare the kinematic recordings of a frequently used field system (physiometer) with two laboratory-based systems (Fastrak and Peak) *in vivo*. Head, thorax and right arm kinematics were recorded simultaneously by the three systems whilst a subject performed 27 single and multiple plane physiological and simulated daily living task movement trials. Errors observed in the Fastrak and Peak data included gimbal lock and quadrant errors. Physiometer data errors included undervalues, overvalues and temporal errors of slow response and resonance. All three systems showed some cross-talk. Agreement between the physiometer and the other systems was generally high for physiological movements ($R^2 > 0.8$) and less for functional movements ($R^2 > 0.5$).

Statement of Relevance: The physiometer recording device can provide an indication of posture across time in the workplace; however, its accuracy is limited, particularly during functional movements. Further technology should be developed to unobtrusively capture accurate all day 3-D kinematics.

Keywords: field based; *in vivo*; motion analysis; posture assessment; validity

1. Introduction

Posture and movement (posture over time) are thought to be important risk factors for the development of work-related musculoskeletal disorders (Gerr *et al.* 1991, Winkel and Westgaard 1992, Kilbom 1994a, Burdorf *et al.* 1997, Li and Buckle 1999, Viera and Kumar 2004). Much of the research on these risk factors has been conducted in laboratories using electrogoniometer (Marras *et al.* 1992, Elford *et al.* 2000), optical capture (Briggs *et al.* 2004) or electromagnetic (O'Sullivan *et al.* 2006) motion analysis systems. Whilst this research has been useful, including providing insights into the physical demands of different tasks and equipment designs, it does not provide adequate information regarding the relationship between overall job exposures and musculoskeletal disorders. Given that the entire exposure of the job is probably important for disorder risk, it is becoming increasingly recognised that all-day posture assessments are necessary (Mathiassen *et al.* 2002, Balogh *et al.* 2004, Viera and Kumar 2004).

Long-term, field based, qualitative motion analyses, such as observation, expert checklists and self-

administered questionnaires, have been popular (Kilbom 1994b, Li and Buckle 1999, Spielholz *et al.* 2001, de Looze *et al.* 2003). Whilst some of these systems have evidence of moderate to high inter-rater reliability (Park *et al.* 2009, Spekle *et al.* 2009), there is also evidence of low accuracy and precision (Wiktorin *et al.* 1993, Hansson *et al.* 2001b). Further, observation methods have been demonstrated to correlate weakly with direct instrumentation (i.e. quantitative measures) (Buchholz *et al.* 2008) and the amount of exposure estimated with different observation indices has also been demonstrated to correlate weakly (van Eerd *et al.* 2009). Therefore, the most recent recommendations for workplace posture and movement assessments are that they should be performed, at minimum, with a combination of direct instrumentation and observation (Teschke *et al.* 2009).

Few all-day posture assessments have been reported to date (Hansson *et al.* 2001b, Balogh *et al.* 2004, Freitag *et al.* 2007), probably due to the difficulties of establishing a suitably accurate, long-term, field-based measurement system. Attempts to

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validate all-day field systems have typically compared output with a laboratory-based motion analysis system. For example, the accuracy of direct field-based measures of lumbar and upper thoracic postures assumed by nurses has been compared with an optical-based laboratory system (Vicon; Oxford Metrics, Inc.) (Morlock *et al.* 2000, Freitag *et al.* 2007). The field-based systems recorded similar flexion angles ($<4^\circ$ of error) as the optical system. However, there was a stark contrast between the angles recorded by the two systems in the frontal and transverse plane angles (Morlock *et al.* 2000, Freitag *et al.* 2007), including reports of the field-based system overestimating these angles by 100% (Morlock *et al.* 2000). This is indicative of 'cross-talk' error, which is particularly evident in some of the common field-based devices.

The 'physiometer' measurement tool has been used for long-term data acquisition of lumbar, neck and humeral posture for two decades (Aaras and Stranden 1988). It is composed of a system of individual inclinometers; 2-D measurement devices based on diversions from gravity. Inclinometers are an efficient, popular method of field-based position recording with varying magnitudes of reported accuracy and reliability (Mayer *et al.* 1984, Miller *et al.* 1992, Rondinelli *et al.* 1992, Levine *et al.* 2005). For instance, the physiometer recorded 3.6° of accuracy when validated on a mechanical rod (Bonato *et al.* 2003); however, faster movement recordings have proved susceptible to phase lag of around 5° (Aaras and Stranden 1988). The physiometer has also been validated *in vivo*, reporting comparable results with 2-D video (Aaras and Stranden 1988, Marras *et al.* 1992). However, 2-D marker digitisation has been linked with significant magnitudes of error and may therefore be an inappropriate 'gold standard' comparison system (Elliott *et al.* 2007). The reliability of inclinometers has been assessed in several investigations. One investigation reported low lumbopelvic position estimation reliability during pelvic tilts, with intra-class correlations (ICCs) of 0.54 recorded when the participant was sitting and 0.36 when they were standing (Henriksen *et al.* 2007). However, when the same device was assessed for cervical position reliability, high inter-assessor (0.89) and intra-assessor (0.93) ICCs were reported (Hoving *et al.* 2005). This unclear efficacy of inclinometers and the physiometer may be clarified by comparisons with a 3-D motion analysis system.

The most accurate quantitative motion analysis systems currently available are laboratory-based systems (Elliott *et al.* 2007). These systems can be used as pseudo 'gold standard' comparison measures for the validation of portable motion recording systems *in vivo* (Mannion and Troke 1999). This is critical, given that the validations on mechanical linkages (any form of

in vitro, non-human device), often reported by manufacturers, have been found to not closely enough replicate 'real' motion analyses. For instance, video-based motion analysis systems, such as the Peak Motus system (Peak Performance Technologies Inc., Centennial, CO, USA), have demonstrated negligible accuracy and reliability errors during planar linkage system motion (Ehara *et al.* 1994, Richards 1999). One investigation reported an accuracy error range from 0 to 1.2° , with a reliability correlation of 0.999 following 10 trial repetitions (Scholz and Milford 1993). However, when analysed on multi-segment, multi-planar linkage system movements, the Peak system consistently recorded errors of around 10° (Elliott *et al.* 2007).

Fastrak (Polhemus Navigation, Kaiser Aerospace, VT, USA) is an electromagnetic device that measures the position and orientation of points in space in three dimensions. Percy and Hindle (1989) recorded an accuracy of 0.2° *in vitro* (Ehara *et al.* 1994, Richards 1999). This system has also demonstrated reliable measures *in vivo* (An *et al.* 1988, Milne *et al.* 1996, Burnett *et al.* 1998); for instance, errors of less than 2% were reported when lumbar spine movements were recorded within 60–70 cm of the source (An *et al.* 1988). This relatively small assessment space is one of the limitations of this system (within 100 cm of the source) (Day *et al.* 1998), together with sensitivity to ferrous objects, which distort the electromagnetic field causing inaccuracies. However, within these constraints, Fastrak can be assumed to be an appropriate 'gold standard' measurement system (Mannion and Troke 1999).

The limited or uncertain accuracy of the currently available direct and indirect all-day posture and motion examinations suggests that previous study conclusions regarding posture as a risk factor for musculoskeletal disorders should be interpreted with some caution. Indeed, the relatively weak and sometimes mixed evidence of risk association may be due to inadequate characterisation of posture and motion. For instance, reports that have found stronger associations between psychological factors and musculoskeletal disorders than physical factors may be the result of inadequate capture of physical exposure estimates (Egger *et al.* 1999, Watson *et al.* 2003).

The lack of standardised collection of posture and motion data further confounds the evidence base. This results from the use of a variety of field-based systems, with goniometers, inclinometers, photographic techniques, electrogoniometers and video recording systems the most frequently reported (Viera and Kumar 2004). Further inconsistencies are introduced through different segment locations and Euler angle conventions. This limits the accurate comparison of

data between investigations and the development of normative databases and indicates that more standardised work posture recording systems and protocols are required. The detailed description of each motion analysis device's potential errors is an essential first step to help researchers select the most appropriate device and assist in the interpretation of previous results.

The aim of this study was to compare the head, thorax and right arm kinematics estimated using a physiometer with those calculated from 3-D laboratory-based systems. A secondary aim was to determine the similarity between 3-D kinematics estimated with video and electromagnetic laboratory systems.

2. Method

This laboratory-based study assessed the motion of one subject simultaneously with three motion analysis systems. Following the provision of information regarding this investigation, consent was obtained. Head, trunk and right arm kinematics were recorded during single and multi-plane physiological movements, as well as simulated work and daily living tasks. The participant was a healthy 44-year-old male subject, with no disorders affecting movement and a height and weight of 175 cm and 72 kg, respectively.

2.1. Motion analysis system's data acquisition

2.1.1. System 1: Physiometer

The physiometer consists of three dual-axis pendulum inclinometers, and was previously described in detail (Aaras and Westgaard 1980). For this study, inclinometers were strapped to the back of the head, on the upper thoracic spine and on the lateral surface of the upper arm, using broad Velcro straps and held in place with additional sports tape (Fixomull[®]; Beiersdorf AG, Hamburg, Germany). Head flexion and lateral bending, thorax flexion and lateral bending and arm flexion and lateral bending (abduction) were sampled at 10 Hz. Segment rotation cannot be recorded with this system. The data were stored temporarily in a portable computer (HP 200LX Palmtop[®]; Hewlett Packard, Palo Alto, CA, USA) worn on the waist.

2.1.2. System 2: 3-Space Fastrak

The 3-space@FastrakTM (Polhemus Navigation Sciences Division, Colchester VT, USA) is an electromagnetic system consisting of a field source and sensors, as previously described (McGill *et al.* 1997). For this study, sensors were placed on the centre of the forehead, over the fifth thoracic spinous

process, and on the right arm using double-sided tape and sports tape. Data were collected at 25 Hz using a customised software program written in LabVIEW V 6.1 (National Instruments, Austin, Texas, USA) on a personal computer. The default angle outputs were calculated using a right-hand coordinate system and a 'ZYX' Euler angle decomposition. The interpretation of the output angles with clinical terminology (flexion, lateral bending and rotation) from the system's nomenclature of roll, elevation, azimuth is detailed in Table 1. This ordered sequence of rotations is the standard recommended by the International Society of Biomechanics (Wu and Cavanagh 1995).

2.1.3. System 3: PEAK

The Peak Motus 8[®] is a near infrared 3-D video-based motion analysis system (Peak Performance Technologies Inc.), as previously described (Briggs *et al.* 2004). Near infrared light is reflected from the passive, retro-reflective markers, such that discrete points are obtained in each video frame. Images from multiple view points are combined to determine points in space using direct linear transformation. For this study, multiple 25 mm diameter reflective markers were used to define head, thorax and right arm segments. Data were sampled at 50 Hz by the Peak Motus[®] 8 (Peak Performance Technologies Inc.) software. Data were automatically digitised and the 3-D points reconstructed; however, no filtering was applied to the data. Angles were calculated using Motus software and the definitions listed in Table 2.

2.2. Movements

Posture was measured whilst the subject performed single plane movements (flexion and lateral bending) of the head, trunk and arm in two ranges – moderate range ($\sim 30^\circ$ for head and trunk and $\sim 60^\circ$ for arm) and large range ($\sim 60^\circ$ for head and trunk and $\sim 120^\circ$ for arm).

Table 1. Clinical angle definition in Fastrak nomenclature.

Angle	System description
Head flexion	Sensor 4 elevation
Head lateral bending	Sensor 4 azimuth
Head rotation	Sensor 4 roll
Thorax flexion	Sensor 3 elevation
Thorax lateral bending	Sensor 3 azimuth
Thorax rotation	Sensor 3 roll
Arm flexion	Sensor 2 elevation
Arm lateral bending	Sensor 2 azimuth

for arm). Movements were performed at a steady pace. Additionally, rapid arm flexion and lateral bending movements were performed (the subject was instructed to move 'quickly'). Multi-plane physiological movements combining flexion, right lateral bending and right rotation were also performed for each body segment. Simulated tasks included office work tasks and activities of daily living. Table 3 details the 27 movement trials performed. Physiological movement trial data were collected during 13 trials of 10 s. One trial each of flexion, lateral bending, rotation and a combination of flexion, right lateral bending and

rotation of the head, trunk and right arm (no rotation) was performed. Office and daily living task trial data were collected during 12 trials of 10 s.

2.3. Procedure

The three measurement systems were simultaneously attached to the subject. Reference postures of the head, right arm and the upper torso were calibrated as the participant stood upright facing forwards, with both arms hanging down by the side of the body. The subject then completed the movement trials.

Table 2. Clinical angle definition in Peak nomenclature.

Angle	Description
Head flexion	This absolute flexion (forward) angle between a vertical vector defined from the Cyclops* and OC1 [†] and a vertical axis 'global' vector.
Head lateral bending	The absolute lateral angle between OC1 [†] , external auditory meatus and vertical axis (negative to the left).
Head rotation	The rotation between a vector defined between OC1 [†] and Cyclops* and the anterior axis (negative to the left).
Thorax flexion	The absolute flexion (forward) angle between a vector from the C7 to the mid trochanter [‡] and a vertical axis 'global' vector.
Thorax lateral bending	The absolute lateral angle between a vector from the T5 and mid trochanter [‡] and a vertical axis 'global' vector.
Thorax rotation	The absolute rotation between a vector from the sternum to T3 and a vertical axis 'global' vector.
Arm flexion	The absolute flexion (forward) angle between a vector from the lateral humeral epicondyle to the acromion and a vertical axis 'global' vector.
Arm lateral bending	The absolute lateral angle between a vector from the lateral humeral epicondyle to the acromion and a vertical axis 'global' vector.

Virtual markers: *mid way between outer canthi; [†]mid way between external auditory meati; [‡]mid way between left and right greater trochanters.

Table 3. Details of each physiological and functional movement performed.

Physiological movements		Functional movements	
Single plane	Combined	Office activities	Activities of daily living
Head flexion to 30° and 60°	Head flexion, right lateral bending and right rotation to 30° and 60° Trunk flexion, right lateral bending and right rotation to 30° and 60° Arm flexion, right lateral bending and counter-clockwise rotation to 60° and 120°	Sitting using keyboard to enter text	Folding washing
Head lateral bending to 30° and 60°		Sitting using mouse to scroll and click and drag	Sweeping
Head rotation to 30° and 60°		Talking on the phone, holding handset to ear	Brushing teeth
Trunk flexion to 30° and 60°		Sitting and writing with a pen and paper	Eating a meal
Trunk lateral bending to 30° and 60°		Reading from a book	Sit to stand to sit
Trunk rotation to 30° and 60°		Talking on phone whilst using mouse, keyboard and pen	Lifting pen from floor
Arm flexion to 60° and 120°			
Arm abduction to 60° and 120°			
Arm flexion to 60° and 120°: fast			
Arm abduction to 60° and 120°: fast			

2.4. Data processing and analysis

Fastrak and Peak data were down-sampled to 10 Hz using cubic spline interpolation using custom LabVIEW software. The three 10 Hz datasets were imported to Excel (Microsoft Corporation, Redmond, WA, USA). Baseline calibration of angle amplitude was performed using the calibration standing data at the beginning of each trial. Automatic temporal synchronisation was not possible during recording so was performed *post hoc* by manual alignment of movement initiation.

Graphs were inspected for data quality and errors such as gimbal lock and quadrant errors (outlined below) were noted and removed with a simple correction factor (e.g. + or -360° in the case of quadrant errors). Cross-talk was investigated by analysing the magnitude of secondary rotation recorded in movements that were performed in a single planar (e.g. during a thorax flexion movement, any recorded 'lateral bending' would be assumed to represent error). For the arm data elevations, a product of flexion and abduction were calculated as per Hansson *et al.* (2001a).

In order to evaluate the systematic and random errors in the physiometer system, independent regression equations were developed comparing the physiometer with 1) the Fastrak and 2) the Peak data, respectively, using the statistical package SPSS version 13 (SPSS Inc., Chicago, IL, USA.). The Fastrak and Peak data were also compared.

3. Results

3.1. Typical system errors

3.1.1. Fastrak and Peak

The main error seen in the Peak data was gimbal lock (82° , 0°) (Figure 1). Gimbal lock occurs when the

Figure 1. Example of gimbal lock error seen in arm flexion to 140° as measured by Peak.

second axis of the 'child segment' in the 'ordered set of rotations' is aligned with the second axis of the 'parent segment'. Therefore, in the case of calculating arm angles with a ZYX ordered set of rotations, the gimbal lock position will occur when the arm (child segment) approaches 90° of rotation relative to the thorax (parent segment). This will result in a reversal of the third angle calculated – in this instance, arm flexion. Fastrak also demonstrated a number of gimbal lock errors; however, the most numerous Fastrak problems were quadrant errors. This is where the data appeared to either add or subtract 360° to the data points, ending up with a graph such as that in Figure 2. Both of these errors were easily detected by visual inspection and could be corrected in Excel.

The Fastrak angles can be output using any of the Euler 'ordered sequence of rotations' to analyse data. This allows researchers to select the ordered sequence of rotations that output the main variable of interest first. However, the biomechanical standard is the clinically interpretable flexion–extension, abduction–adduction and internal–external rotation (Wu *et al.* 2005). The Fastrak data were output with a ZYX ordered sequence of rotations, which followed the standard flexion/abduction/rotation interpretation. In some instances, particularly the arm abduction data, this appeared to be the incorrect sequence to use, as indicated by the poor R^2 values. It was decided to re-analyse some of the arm data to quantify the errors produced by a poor sequence choice. As can be seen in Figure 3a, using the sequence XYZ (abduction, flexion, rotation) does not change the primary (abduction) angle when performing a single plane movement, but does change the secondary (flexion) angle (Figure 3b).

3.1.2. The physiometer

The physiometer data undervalued the range of motion performed during the 60° thorax flexion, 120°

Figure 2. Example of quadrant error seen in head rotation to 30° recorded by Fastrak.

arm abduction, 120° arm flexion (Figure 4) movements, as well as the 60° combination (multiplanar) movement (only in arm abduction). For instance, the physiometer recorded a maximum of 51° flexion (compared with Fastrak 69° and Peak 65°) and a minimum lateral movement of -70° (compared with Fastrak -141° and Peak -122°), during the 60° combined thorax movement. This was partly due to the data 'flip', when the inclinometer exceeded 73°. The physiometer overvalued flexion range of movement in 60° combined head movement, recording a maximum of 96°, compared to the 56° recorded by Fastrak and 73° recorded by Peak. The physiometer also demonstrated temporal errors. For example, during the 60° combined arm movement the physiometer reached a plateau of 85°, the same

maximum value as Fastrak and Peak; however, it was reached and held for 0.8 s before and 0.6 s after Fastrak and Peak data.

The physiometer also demonstrated a slow response during rapid movements. As demonstrated in Figure 5 during an arm movement trial, the physiometer shows close alignment with the Peak and Fastrak angles during the execution of a 'slow' motion. However, during the fast (second) movement, the physiometer demonstrates little agreement with the movement curves collected with the Fastrak and Peak systems. Resonance occurred in the arm flexion physiometer data during the trial of teeth brushing (rapid oscillating movement), as illustrated in Figure 6.

Whilst all systems demonstrated some cross-talk error during the single plane trials, this error was most

Figure 3. (a) Example of minimal effect of reordering Fastrak sequence of rotations on primary angle (arm abduction) during arm abduction to 60° movement (the recalculated Fastrak XYZ line is hidden under the original Fastrak line); (b) example of the effect of reordering Fastrak sequence of rotations on secondary angle (arm flexion) data during the arm abduction to 60° movement.

Figure 4. Example of undervalue and flip errors in physiometer data during arm 150° flexion movement.

Figure 5. Example of slow response in physiometer data during fast arm flexion to 60° movement (the first movement is at a normal speed, the second movement is at a fast speed).

evident in the angles calculated with the physiometer. For instance, in Figure 7, arm flexion at 60° is represented by the thicker line and the cross-talk is shown in arm abduction. All three systems showed some cross-talk in this trial; however, the physiometer error was largest.

3.2. Agreement between systems

In single plane physiological movements R^2 ranged from 0.436 (Peak vs. physiometer in 60° arm abduction) to 0.999 (Fastrak vs. Peak in head flexion to 30°). Of the 43 comparisons 39 (90.7%) were over 0.8. All 17 of Peak vs. Fastrak comparisons were over 0.8, and only one was below 0.9. The poorest R^2 were seen in

Figure 6. Example of resonance error oscillations in physiometer arm flexion data during brushing teeth movement.

Figure 7. Example of cross-talk error in arm abduction recorded by each system during an arm flexion movement. The thicker line is the actual arm flexion movement; the other lines are measures of adduction, largely the result of cross-talk.

the physiometer comparisons during 120° arm abduction (Fastrak vs. physiometer, 0.487; Peak vs. physiometer, 0.436) and the fast arm flexion (Fastrak vs. physiometer, 0.656; Peak vs. physiometer, 0.674). Table 3 summarises mean R^2 for single plane head, thorax and arm movements. The R^2 means for all single plane comparisons were 0.901 for physiometer vs. Fastrak, 0.910 for physiometer vs. Peak and 0.980 for Fastrak vs. Peak.

Table 3 also summarises mean R^2 for combined plane movements. In multiple plane physiological movements R^2 ranged from 0.061 (Peak vs. physiometer in lateral bending during head combination to 30°) to 0.998 (Fastrak vs. physiometer in flexion during head combination to 30°). In total, 30 of the 37 comparisons (81.1%) were over 0.8. The poorest R^2 were again seen with physiometer comparisons, this time for lateral bending during the head combination to 30° movement (Fastrak vs. physiometer, 0.152; Peak vs. physiometer, 0.061; Fastrak vs. Peak, 0.369) and arm abduction during the arm 120° combination movement (Fastrak vs. physiometer, 0.663; Peak vs. physiometer, 0.895; Fastrak vs. Peak, 0.485).

Table 4 also summarises mean R^2 for office and daily living movements. In office and daily living movements R^2 ranged from 0.000 (Peak vs. physiometer in head lateral bending during pen to floor task) to 0.996 (Peak vs. Fastrak in head flexion during pen to floor task). In total, 84 of the 240 comparisons (70%) made were over 0.8. As would be expected, the poorest R^2 were seen in activities with very low movement ranges, such as thoracic lateral bending during the keyboard task (Figure 8). Table 5 shows that across all systems and both single and combined plane movements, R^2 tended to be less for the large range movements.

Table 4. Mean regression coefficients (R^2) for physiometer, Peak and Fastrak comparisons for physiological and functional movements.

	Physiometer vs. Fastrak	Physiometer vs. Peak	Fastrak vs. Peak
Physiological			
Head	0.972	0.983	0.996
Thorax	0.963	0.986	0.982
Arm	0.794	0.790	0.958
Combined movement	0.878	0.870	0.854
Functional			
Office activities	0.525	0.550	0.611
Daily living activities	0.596	0.481	0.588

Figure 8. Thorax lateral bending whilst using a keyboard.

Table 5. Mean regression coefficients (R^2) for physiometer, Peak and Fastrak comparisons for moderate and large range movements.

		Moderate range	Large range
Head	Flexion	0.992	0.956
	Lateral bending	0.992	0.994
Thorax	Flexion	0.985	0.937
	Lateral bending	0.994	0.989
Arm	Flexion	0.985	0.896
	Lateral bending	0.994	0.593

3.2.1. Using elevations to improve agreement between systems

A method of reducing cross-talk error for arm movements in the physiometer has been proposed by Hansson *et al.* (2001a). It involves the conversion of abduction and flexion angles into a single projection angle of elevation. Table 6 shows that the R^2 were generally improved using this correction.

4. Discussion

Accurate all-day acquisition of posture and movement data is critical for the investigation of this as a risk factor for musculoskeletal disorders. The physiometer has proved to be a popular method of field-based posture estimation (Aaras and Stranden 1988, Ciccarilli *et al.* 2007). However, the previous *in vivo* physiometer validation process (Aaras and Stranden 1988) may have been confounded by the use of 2-D video as a criterion. Therefore, this study aimed to compare the head, arm and thorax angles recorded by the physiometer with two laboratory-based 3-D systems (Fastrak and Peak).

3-D laboratory-based systems are frequently used as criterion measures for the validation of kinematic recording equipment (Morlock *et al.* 2000, Elliott *et al.*

Table 6. Comparison of regression coefficients (R^2) for arm flexion, lateral bending and elevation for the large range arm flexion movement.

	Physiometer vs. Fastrak	Physiometer vs. Peak	Fastrak vs. Peak
Flexion	0.845	0.853	0.896
Lateral bending	0.487	0.436	0.593
Elevation	0.927	0.951	0.935

2007, Freitag *et al.* 2007). However, every measurement system will be associated with some magnitude of error. These errors can be categorised as 'instrumental' errors, which can largely be removed with data treatment, and 'real errors', which occur from soft tissue movement of the surface markers that is not representative of underlying skeletal motion. The most accurate lumbar measurements will be direct measures, such as the application of Steinmann pins into the spinous processes of the lumbar vertebrae (Pope *et al.* 1992) or the use of dynamic magnetic resonance imaging (MRI) to track skeletal motion. However, even these techniques will be associated with some imprecision and are not suited to field studies.

The main errors associated with Fastrak and Peak systems were gimbal lock and quadrant errors. These result from mathematical inconsistencies unrelated to the system specifications (i.e. errors that will be evident in all 3-D systems). Gimbal lock errors occur when the axes align during the calculations. This results in a 'zero' value and incorrect re-orientation (or 'flipping') of the data. The quadrant errors occur when the entire orientation of the 3-D system is confused, resulting in a precise 360° error in the angle calculation. Both of these errors require minimal data treatment to correct and can therefore be considered as a treatable 'side effect' of 3-D data representation. The Euler ordered sequence of rotations used to calculate angles can also be configured to minimise such errors. For instance, they can be manipulated such that the gimbal lock error will occur in a limb location that is anatomically impossible to reach; for instance, at 90° of elbow abduction. The ordered sequence of rotations can also be changed to alter the 'cross-talk' error. However, the manipulation of the outputs in this manner will affect the interpretation of the angles and it is therefore important that the biomechanical standard for all joints is followed where possible.

The agreement between Fastrak and Peak systems during single plane movements was quite high (minimum correlation of 0.8). However, there was relatively low agreement between the two systems when used to record multi-plane movements. For example, the average correlation between Fastrak and Peak for all recorded activities of daily living was only

0.588. This pattern, where strong agreement is found in single plane motion and limited agreement, is found in multi-plane movements, is similar to results previously documented by Elliott *et al.* (2007) and further highlights the importance of using multi-plane movements when validating measurement equipment. Therefore, accuracy specifications typically reported by equipment manufacturers, while a necessary indication of the system's precision, cannot be interpreted as representative of 'practical accuracy'. The errors associated with the Peak system during multi-plane movements are likely the result of incorrect marker digitisation and that the Peak is not a truly 3-D system. Therefore, inaccuracies associated with making largely planar analyses on data of multi-plane movements can occur (Elliott *et al.* 2007).

The physiometer demonstrated variations from the data recorded with the other two systems in most of the movements. The physiometer angle calculation errors are largely the result of the hardware composition of the inclinometers. For instance, oscillations not representative of the relatively smooth movements performed were evident, particularly in the small range, rapid movements, such as during the 'brushing teeth' activity. This can be linked with the 'pendulum' of the inclinometers swinging. The R^2 were reduced by these oscillations (Fastrak vs. physiometer 0.847; Peak vs. physiometer 0.819 Fastrak and Peak comparison (0.973)); however, this error could be partly corrected with a filtering routine. The more concerning errors within the physiometer data were those that cannot be easily corrected. These include the inability to record the full range of movements when large ranges were performed ($> 60^\circ$), as well as the 'phase leg' evident when the 'pendulum' type measurement device miscalculated rapid motion, resulting in the physiometer not detecting a large proportion of the motion recorded by the Fastrak and Peak systems during fast movements.

All the systems investigated recorded some magnitude of cross-talk error. It is relatively impossible to perfectly align non-invasive measuring devices with the underlying skeleton and axes of rotation. Therefore, in any angle measure, where the movement in another plane is recorded, some 'cross-talk' in the incorrect plane will likely occur. However, this error can be minimised by the careful alignment of surface devices. 2-D systems will be more susceptible than 3-D systems to cross-talk error when used to record 3 degrees of freedom (DOF) movements. This indicates that the physiometer will be more susceptible than both the Fastrak and Peak and was supported by the results of this investigation. The physiometer demonstrated considerably greater cross-talk than both Fastrak and Peak. This error cannot be easily corrected with conventional data reduction methods, such as data

smoothing. However, the elevation correction outlined by Hansson *et al.* (2001a) resulted in a significant improvement in the agreement between the physiometer and the other two systems. Therefore, this correction process can be considered a significant enhancement to the physiometer's accuracy, at the expense of losing separate flexion and adduction measures.

The results of this investigation suggest that the physiometer is a relatively useful recording device during slow to moderate pace and moderate range movements, particularly at 2 DOF joints. However, for the purpose of all-day assessment of lumbar, thorax, neck and upper arm postures it has more limited accuracy, given that rapid, wide-ranging, multi-plane movements are expected to occur.

This investigation compared angle recordings of three measurement devices during a number of multi- and single plane tasks, performed at varied ranges of speed and motion, on multiple segments. This protocol is necessary to clarify the differences between systems, given that comparisons on single plane movements and *in vitro* devices have been demonstrated as potentially misleading (Elliott *et al.* 2007). The inclusion of three systems allowed a more detailed analysis and provided an estimation of the most accurate system. The results documented herein highlight the potential imprecision associated with every recording device.

The limitations of this study include an inability to place all three systems on identical bony landmarks due to space restrictions. For instance, the Peak thorax data may have been confounded by the use of a marker location unique to the other two systems. This marker location may have been subject to different magnitudes of 'soft tissue movement' related errors, which could not be recorded in this investigation, which compared only non-invasive measurement systems. A further limitation was the use of only one subject. Different anthropometry may be associated with variations in error, especially surface-bone error in obese individuals. Other subject-specific characteristics may also have influenced the results. The down-sampling of Fastrak (25 Hz) and Peak (50 Hz) sampling frequencies to match the physiometer frequency (10 Hz) may have resulted in the misrepresentation of some of the original angles calculated. However, this is not likely in the slow velocity movements, with only a minimal effect expected in the angles calculated during the fast movements. Further, as no direct measures were recorded, e.g. Steinmann pins or dynamic MRI, there is no way of distinguishing between the accuracy of the different recordings. Assumptions made regarding the 'most accurate' systems therefore rely on previous research, the amount of necessary data treatment required and knowledge of the movement performed.

4.1. Future options

A comprehensive measurement strategy is paramount to gain accurate estimates of risk factor exposures for work-related musculoskeletal disorders (Burdorf 1992). In light of the results of the present study and the limitations of the physiometer, other systems should be assessed for both efficacy and suitability for all-day acquisition.

Electrogoniometers record 2 DOF movements through two piezoelectric strain gauge devices connected with a measuring cable (Legnani *et al.* 2000). The *in vivo* accuracy of these systems correlates strongly with 3-D video data (0.84–0.97) (Brumagne *et al.* 1999) during single plane motion. However, as previously mentioned, 3-D video data are not a suitable criterion measure and multi-plane validations should be performed. Electrogoniometers have been linked with several limitations. The ‘lumbar motion monitor’ records posture through a system of electrogoniometers (Marras *et al.* 1992). However, its cumbersome, obtrusive design may alter natural spinal postures and would not be suitable for assessing lumbar posture when using conventional office chairs. Small flexible strain gauges of commercial electrogoniometers are less intrusive but may be susceptible to bending or buckling during studies of seated tasks with lumbar support (Boocock *et al.* 1994). The measurement range of the spinal motion could be limited by exceeding the maximum preset distance between the two attachments with linear displacements occurring during flexion and extension of the lumbar spine. Susceptibility to cross-talk errors has also been reported; however, these errors ranged between 1° and 3°, a margin of error that might be disregarded in a number of ergonomic and epidemiological applications (Hansson *et al.* 2004). Moreover, following the addition of a torsionmeter, which can record lumbar rotations, cross-talk correction algorithms can be used to further minimise this error (Hansson *et al.* 2004). The electrogoniometer’s light/flat design is relatively non-intrusive and it can collect for long periods of time (around 12 h). The *in vivo* validation of the combined electrogoniometer/torsionmeter system to record multi-plane movements is therefore required.

Inclinometers such as those utilised in the physiometer are popular, lightweight recording devices, with several accuracy limitations. The intelligent device for energy expenditure and activity (IDEAA) is a currently available, accurate measurement tool for determining the frequency, type and intensity of daily movements (Zhang *et al.* 2003). This technology was developed to monitor the energy associated with movements. For the purpose of movement detection, the IDEAA has proved accurate, correctly identifying gross posture

and limb movement type on average for 98.9% of the time and the gait type on average for 98.5% of the time. While the angle sensors have been documented to provide accurate estimations of the type of movement performed, the precision of these devices to record the magnitude of motion has not been verified. The system can operate for up to 48 h (Zhang *et al.* 2003).

Accelerometers detect the acceleration of a moving body. These have proved popular in a number of applications, including energy expenditure estimation (Levine and Miller 2007). The acceleration data can also be ‘double integrated’ to determine position; however, integration drift errors are a significant issue, resulting in errors of up to 1 m (Willemssen *et al.* 1991, Roetenberg *et al.* 2007). Gyroscopes estimate the rate of positional change or angular velocity between segments. They have been found to be reasonably reliable for this purpose. However, when the angular velocity information is integrated to calculate position, the same ‘integration drift’ errors experienced by accelerometers may occur (Roetenberg *et al.* 2007). Recent advances in sensor integration have led to the development of inertial sensors, which combine accelerometer and gyroscopes with magnetometers to overcome the integration drift error. They have been implemented in a number of investigations, including the assessment of daily humeral postures (Zhou and Huosheng 2004, Coley *et al.* 2008), spinal posture assessments (Goodvin *et al.* 2006) and gait analyses (Roetenberg *et al.* 2007). These systems use the global reference position estimated with magnetometers to correct for integration drift associated with gyroscopes and accelerometers. The data supplied by the magnetometer, accelerometer and gyroscope is combined with sensor-fusing algorithms. Inertial sensors are lightweight and, whilst current commercially available systems have a limited battery life (3 h), batteries can be changed without disturbing the collection signal. The functional accuracy in comparison with an optoelectronic system has been demonstrated (Cutti *et al.* 2008), with a maximum error of 3.6° when used to estimate shoulder and elbow kinematics, which suggests an acceptable accuracy. A potential limitation associated with these sensors is the cube-like dimensions (height of 2.8 cm, Xsens) (Cutti *et al.* 2008). If located on the lumbar vertebrae and sacrum, this amount of protrusion will likely impede normal chair-sitting posture.

Markerless video tracking is an active and promising but also challenging research area that has emerged in the last decade (Zhou and Huosheng 2004). There are many unresolved technical problems, with active shape detection and contour mapping models currently offering guides to the body task being undertaken. Several of the 2-D methods offer relatively

simple methods of detecting the type of movement or the behaviour being performed (Long and Yang 1991, Wren *et al.* 1997). However, the more accurate 3-D methods, whilst offering estimates of body posture, are not yet accurate enough for most ergonomic purposes and require the use of a number of complex algorithms that are associated with long computation time and generally rely on complex kinematic restraints (Barron and Kakadiaris 2003, Zhou and Huosheng 2004).

A further interesting issue is that the required accuracy of any system to determine long-term musculoskeletal disorder risk is unknown. This is partly because accurate field-based systems have not been available so that the relationship between long-term posture/movement and musculoskeletal disorder is underspecified.

4.2. Conclusion

Work-related musculoskeletal disorders are a significant burden to health care and industry. Efforts to understand the contribution of posture and movement to these disorders has been hampered by limitations of exposure assessment methods. A number of different systems are currently in use, with varying levels of accuracy and reliability often verified thorough unsophisticated, *in vitro*, validations. Several of the currently available systems, such as the physiometer, have demonstrated an ability to provide some useful information that is more precise than self-report or structured observation techniques. However, they are limited in accuracy and time-consuming in acquiring data. Therefore, a number of technological approaches are being pursued in the hope of developing more precise, efficient instrumentation. To date, no unobtrusive system is readily available. In the near term, inertial sensors may be useful for limb movement assessments, where the cube-like dimensions are acceptable, and electrogoniometers, combined with torsionmeters for 3 DOF data acquisition and cross-talk minimisation, may be useful where the inertial sensor size is too obtrusive, such as sitting lumbar posture examinations.

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