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APPLICATION OF INERTIAL MEASUREMENT UNITS FOR DIRECTLY MEASURING OCCUPATIONAL EXPOSURE TO NON-NEUTRAL POSTURES OF THE LOW BACK AND SHOULDER

by

Mark Christopher Schall, Jr.

A thesis submitted in partial fulfillment
of the requirements for the Doctor of
Philosophy degree in Industrial Engineering
in the Graduate College of
The University of Iowa

December 2014

Thesis Supervisors: Assistant Professor Nathan Fethke
Associate Professor Geb Thomas

Graduate College
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CERTIFICATE OF APPROVAL

PH.D. THESIS

This is to certify that the Ph.D. thesis of

Mark Christopher Schall, Jr.

has been approved by the Examining Committee
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ABSTRACT

Epidemiological evidence suggests an association between exposure to non-neutral working postures and work-related musculoskeletal disorders (MSDs) of the low back and shoulder. Accurate and precise quantitative estimation of exposure to non-neutral working postures is, therefore, essential for evaluating worker risk, developing and testing ergonomic interventions, and improving worker health and well-being. Current methods used to directly estimate occupational exposure to non-neutral postures may be obtrusive, often lack sufficient portability for field use, and have limited accuracy and precision when used to measure dynamic or complex motions.

Inertial measurement units (IMUs) are emerging instrumentation devices that measure and report an object's orientation and motion characteristics using multiple electromechanical sensors (i.e., accelerometers, gyroscopes, and/or magnetometers). They have been observed to accurately monitor body kinematics over periods of relatively short duration in comparison to laboratory-based optical motion capture systems. Limited research, however, has been performed comparing exposure information obtained with IMUs to exposure information obtained with other field-capable direct measurement exposure assessment methods. Furthermore, insufficient information on the repeatability of IMU-based estimates over a substantial time period (e.g., a full work shift) and inadequate knowledge regarding the effects of different IMU sensor configurations and processing methods on the accuracy and repeatability of estimates of exposure obtained with IMU systems contributes to a lack of their use in epidemiological field studies.

This thesis was designed to address these issues and expand upon the current scientific literature regarding the use of IMU sensors as direct measurement devices for assessing exposure to non-neutral working postures in the field. Chapter I provides a background and justification for the work. Chapter II presents the findings of a laboratory-based, manual material handling study that was performed to compare

estimates of thoracolumbar trunk motion obtained with a commercially available IMU system with estimates of thoracolumbar trunk motion obtained with a field-capable reference system, the Lumbar Motion Monitor (LMM). The effects of alternative sensor configurations and processing methods on the agreement between LMM and IMU-based estimates of trunk motion were also explored. Chapter III presents the results of a study performed to evaluate the accuracy and repeatability of estimates of trunk angular displacement and upper arm elevation obtained with the IMU system examined in Chapter II over the course of an eight-hour work shift in both a laboratory and field-based setting. The effects of alternative sensor configurations and processing methods on the accuracy and repeatability of estimates of trunk angular displacement and upper arm elevation obtained with the IMU system were also studied. Chapter IV presents the results of a randomized, repeated measures intervention that demonstrates the utility of the IMU system examined in Chapters II and III as a direct measurement instrument for comparing “ergonomic” and conventional examination equipment commonly used by ophthalmologists. Finally, Chapter V summarizes the major findings, discusses their practical implications, and provides suggestions for future research.

PUBLIC ABSTRACT

Low back pain and disorders of the shoulder are among the most common and expensive of all occupational injuries and illnesses. Gaining a better understanding of how and why workers develop these conditions is important for protecting worker safety, health, and well-being. Current methods used to measure occupational exposure to non-neutral working postures, a common risk factor associated with low back pain and shoulder disorders, in a real work environment are lacking. Novel technology has recently become available that may be better suited for measuring occupational exposure to non-neutral working postures than previous methods. However, limited research has been conducted to evaluate this technology for use in real work environments.

The aims of this thesis were, therefore, to (i) evaluate the accuracy and repeatability of the novel technology designed to estimate exposure to non-neutral working postures, (ii) explore the effects of wearing the technology in slightly different configurations and using different data processing options on its accuracy and repeatability, and to (iii) demonstrate the effectiveness of the technology by applying it in a study comparing old work equipment with innovative work equipment designed to reduce the development of disabling low back and shoulder conditions in healthcare workers. Results of the thesis showed that the novel technology may be used to accurately and repeatedly estimate exposures to non-neutral working postures in studies involving real work environments. Researchers can now more confidently use this technology in studies designed to better understand the reasons why people develop disabling low back and shoulder conditions.

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Schall Jr., M.C., Fethke, N.B., Chen, H., Gerr, F. (under peer review - revision submitted to journal on 10.29.2014). A comparison of instrumentation methods to estimate thoracolumbar motion in field-based occupational studies. *Applied Ergonomics*.

Chapter III

Schall Jr., M.C., Fethke, N.B., Douphrate, D., Oyama, S., Chen, H. (In preparation). Accuracy and repeatability of an inertial measurement unit system for field-based occupational studies.

Chapter IV

Schall Jr., M.C., Fethke, N.B., Chen, H., Kitzmann, A. (2014). A comparison of examination equipment used during common clinical ophthalmologic tasks. *IIE Transactions on Occupational Ergonomics and Human Factors*, 2 (2), 105-117.

CHAPTER I

BACKGROUND AND SIGNIFICANCE

Work-related Musculoskeletal Disorders

The National Institute for Occupational Safety and Health (NIOSH) recognizes work-related musculoskeletal disorders (MSDs) as a major cause of illness and disability among workers in many occupations (NIOSH, 1997; National Research Council and the Institute of Medicine [NRC-IOM], 2001). The term MSD refers to a wide range of inflammatory and degenerative conditions affecting the muscles, tendons, nerves, ligaments, joints, or spinal discs (Punnett, 2014). Often characterized as adverse health outcomes of non-traumatic actions repeated over an extended period of time (McCauley-Bush, 2011), MSDs may fluctuate in severity from mild periodic symptoms to severe chronic and debilitating conditions (Piedrahíta, 2006). Some well-documented examples of MSDs include carpal tunnel syndrome, tendonitis of the hand and wrist, shoulder impingement syndrome, and low back pain (LBP).

According to the United States Bureau of Labor Statistics (BLS), there were roughly 1.2 million reported cases of nonfatal occupational injuries and illnesses that required days away from work to recuperate in 2012 for private industry, state government, and local government employees. MSDs accounted for 34% of these workplace injuries and illnesses (BLS, 2013), suggesting they are the leading cause of disability and morbidity in working people in the US. The high rate of MSDs among working people in the US has been a consistent issue over the past two decades, with the disorders annually representing between 29% and 35% of all occupational injuries and illnesses involving days away from work since 1992 (American Federation of Labor and Congress of Industrial Organizations [AFL-CIO], 2012).

The burden of MSDs is not unique to the United States. MSDs were estimated to account for 31% of all occupational diseases in the world in 1994, suggesting that they were the most frequently occurring disease affecting workers at that time (Leigh et al., 1999). Relatively new estimates suggest that MSDs remain the second greatest cause of disability and have the fourth greatest impact on the health of the world population, when considering both death and disability (Murray et al., 2013). The high prevalence of MSDs is only expected to increase as people live longer, work into older age more frequently, and medical conditions such as obesity continue to proliferate (Cavuoto and Nussbaum, 2014; Woolf and Pfleger, 2003).

The predominance of MSDs results in a substantial economic burden that is difficult to estimate precisely. In 1996, NIOSH conservatively estimated that MSDs cost a total of $13 billion annually. In comparison, Praemer et al. (1999) estimated that $215 billion was spent on direct and indirect costs associated with work-related and non-work-related MSDs in 1995 alone. In 2001, the NRC-IOM estimated that the total cost associated with reported MSDs was between $13–54 billion, or roughly 0.8% of the US's Gross Domestic Product (GDP). Most recently, direct and indirect costs for MSDs were estimated to total $1.5 and $1.1 billion using worker compensation medical and indemnity costs from the state of Ohio (Bhattacharya, 2014). Regardless of the estimate, it is evident that MSDs pose an important problem in both health and economic terms. Furthermore, considerable underreporting likely leads to an underestimation of their true burden (Baldwin, 2004; Boden and Ozonoff, 2008; Morse et al., 2005).

Conditions of the low back and shoulder are among the most common and costly of all MSDs. Lifetime prevalence estimates of LBP and shoulder complaints in the general population have been estimated to reach 84% and 66.7%, respectively (Airaksinen et al. 2006; Luime et al., 2004; Walker 2000; Rubin, 2007). The prevalence of LBP has been observed to be the highest during middle age, which represents the largest proportion of a person's working life (Hoy et al., 2012). The prevalence of shoulder pain in the working population has been estimated to be 12% (Miranda et al. 2005), but is estimated to be greater in some occupations (Leclerc et al., 2004;

Occhionero et al., 2014). Of the 1.2 million reported cases of nonfatal occupational injuries and illnesses that required days away from work to recuperate in 2012, the back was injured in nearly half of the MSD cases and required a median of 7 days to return to work. MSDs of the shoulder were observed to be more severe as they required a median of 24 days away from work (BLS, 2013). Total health care expenditures incurred by individuals with LBP in the United States have been estimated to total $90.7 billion (Luo et al., 2004), while the median cost per claim has been estimated to total $8,750 for the low back and $6,668 for the shoulder (Dunning et al., 2010).

Several risk factors have been associated with the development of MSDs including personal characteristics (e.g., age, gender, body mass index, previous musculoskeletal conditions), exposure to physical risk factors (e.g., forceful muscular exertions, repetitive motions, vibration), psychosocial stress (e.g., high job demands and low job control), and workplace organizational factors (e.g., absence of job rotation strategies) (David, 2005; Gerr et al., 2014a, 2014b; Manchikanti, 2000; van der Windt et al., 2000). Although not exclusively caused by exposure to physical risk factors at work, occupational exposure to physical risk factors has been established as a chief risk determinant of MSDs (David, 2005; Gerr et al., 2014a; Tanaka et al., 2001; Winkel and Westgaard, 2000). Several critical systematic reviews associating exposure to physical risk factors at work have suggested a causal relationship exists between MSDs of the low back and shoulder and exposure to non-neutral working postures (da Costa and Vieira, 2010; Hagberg and Wegman, 1987; Jonsson, 1988; Miranda et al., 2001; Ohlsson et al., 1994; Punnett et al., 1991, 2000; NIOSH, 1997; Silverstein et al., 2006, 2008; Svendsen et al., 2004a, 2004b; Vandergrift et al., 2012; van der Windt et al., 2000; Viikari-Juntura et al., 2001). This thesis will examine the physical risk factor of *non-neutral postures*, with particular focus on the low back and shoulder.

Non-neutral Postures as a Physical Risk Factor for

Musculoskeletal Disorders of the Low Back and Shoulder

Non-neutral postures generally refer to positions of the body that deviate substantially from a relaxed or resting position. When a non-neutral posture is assumed, some of the body's muscles are forced to provide supplementary support to compensate for the uneven weight distribution created by the non-neutral position. This compensatory muscle activity places the body's muscles, tendons, ligaments, and joints at risk for overloading and injury (NIOSH, 1997). Risk for injury is further exacerbated when the non-neutral postures are held continuously (often referred to as static postures), are performed repeatedly, or are accompanied by forceful muscle exertions (Delleman et al., 2004; NIOSH, 1997; Vieria and Kumar, 2004).

Despite agreement among the majority of the ergonomics community that working in non-neutral postures poses a risk for MSDs, the exact amount of deviation necessary to constitute a substantial deviation from neutral remains ambiguous. One potential explanation for the ambiguity is that associations between exposure to non-neutral working postures and musculoskeletal outcomes of the low back and shoulder have been observed in studies using inconsistent summary metrics (Palmer et al., 2012; Punnett and Wegman, 2004). For example, many occupational studies of trunk and shoulder posture have commonly used percentiles of the cumulative posture distribution such as the 10^{th} and 90^{th} percentiles to assess the occurrence of neutral and extreme postures (e.g., Delisle et al., 2006; Kazmierczak et al., 2005; Wahlstrom et al., 2010). Variables based on percentiles, however, have been observed to suffer from a lack of precision in samples of limited duration (Mathiassen et al., 2012). For this reason, alternative summary metrics such as the proportion of time spent in pre-defined angle sectors may be preferred.

Non-neutral Posture Definitions for the Trunk and Low Back

Body positions that constitute non-neutral postures of the trunk and low back have generally been defined as any “extreme deviation” from the neutral position in up to six possible directions: forward flexion or backward extension in the sagittal plane, lateral bending to the left or right in the coronal plane, or axial rotation (twisting) to the left or right in the transverse plane (NIOSH, 1997). Definitions of neutral trunk postures in recent epidemiological studies have included working with the trunk flexed in angles <20°-30°, while extreme or “severe” postures have been generally expressed as angles ≥60° (NIOSH, 1997; Kazmierczak et al., 2005; Fethke et al., 2011; Hooftman et al., 2009). It has recently been recommended that forward flexion or backward extension in the sagittal plane and trunk lateral bending to the left or right in the coronal plane be partitioned into 4 categories of 30° increments and 3 categories of 15° increments, respectively (NIOSH, 2014).

While such angle categories have been used in an attempt to more precisely characterize associations between non-neutral working postures and MSDs of the low back, evidence suggests that MSDs can occur across a range of motions. For example, Hoogendorn et al. (2000) observed an increased risk of LBP for workers who worked with the trunk flexed a minimum of 60° for >5% of the working time (RR 1.5, 95% CI 1.0 – 2.1). In a case-control investigation of auto assembly workers, Punnett et al. (1991) observed a positive association between MSDs of the back and the percentage of work time with the trunk flexed 20º - 45 º (OR 4.9, 95% CI 1.4 – 17.4). The risk for LBP was elevated (OR 5.7, 95% CI 1.6 – 20.4) when workers were exposed to trunk flexion angles >45º or to two or three non-neutral postures in a work cycle. Jansen et al. (2004) reported a similar risk of disabling LBP among nursing home employees when occupational exposure to trunk flexion exceeded 45º.

Other studies examining associations between physical risk factors and several low back musculoskeletal outcomes such as Seidler et al. (2001) have observed the cumulative number of hours spent in trunk flexion >90° was associated with LBP-related diagnoses among patients from a variety of occupations. In a study designed to examine exposure prediction models for mean, peak (i.e., 90th percentile), and cumulative low back muscle activity, trunk flexion angles exceeding 60º were observed to be positively related to both mean and peak muscle activity readings, suggesting that trunk flexion angles >60 º may be biomechanically meaningful (Trask et al., 2010). However, this interpretation may not take into account the flexion-relaxation phenomenon that occurs in healthy individuals that may influence EMG measurements of the back muscles (Solomonow et al., 2003).

Non-neutral Posture Definitions of the Shoulder

The shoulder joint is one of the largest and most complex joints in the human body. Comprised of three bones (the clavicle, humerus, and scapula) and a collection of muscles, tendons, ligaments, and various protective tissues, the shoulder allows the upper arm to abduct, adduct, rotate, and move through a full 360° in the sagittal plane. This extensive range of motion makes the shoulder prone to musculoskeletal pain and injury.

Definitions of neutral shoulder posture in the scientific literature have included working with the upper arms flexed or abducted in angles <20° (Kazmierczak et al., 2005; Wahlstrom et al., 2010; Bao et al., 2009), <30° (Hooftman et al., 2009, Juul-Kristensen et al., 2001) and 45° (Keyserling, 1986), and extreme or “severe” postures have been expressed as angles ≥60° (NIOSH, 1997; Kazmierczak et al., 2005; Wahlstrom et al., 2010; Hooftman et al., 2009; Hansson et al., 2006) or ≥90° (Svendsen et al. 2004a; Keyserling, 1986). NIOSH has recently suggested shoulder abduction range of motion be partitioned into 5 categories of 30° (NIOSH, 2014).

Similar to the low back, evidence suggests that MSDs can occur across a range of upper arm motions. Silverstein et al. (2008), for example, identified increased odds of developing rotator cuff syndrome when arm flexion exceeded 45° in a cross-sectional study of 733 manufacturing and healthcare sector workers in Washington State. In a study aimed at determining quantitative exposure-response relationships between work with highly elevated arms and shoulder conditions, Svendson et al. (2004a) were not able to discern between the effects of work with the arm elevated above 60˚and above 90˚, but did provide evidence of a relation between shoulder disorders at both heights. In a systematic review of the literature assessing of the exposure-response relationships between work-related physical and psychosocial factors and the occurrence of specific shoulder disorders in occupational populations, van Rijn et al. (2010) observed working with hand above shoulder level showed an association with shoulder impingement syndrome (OR 1.04 - 4.7) as well as upper-arm flexion >45° for >15% of work time (OR 2.43). More recent work examining associations between physical risk factors and MSD risk suggests that the percentage of time with the shoulders elevated >90° is observed as the physical risk factor most strongly associated with neck/shoulder symptoms and disorders (Gerr et al., 2014a).

Other Factors Contributing to Heterogeneity of Associations

Inconsistent definitions of neutral and extreme trunk and shoulder postures are not the only factor contributing to the heterogeneity of the association between exposure to non-neutral working postures and musculoskeletal outcomes. Other factors include a limited understanding of the pathogenesis of low back and shoulder MSDs, the multifactorial etiology of MSDs, and methodological limitations.

Evidence suggests that the intervertebral disc, and more specifically, discogenic pain as a result of degenerative disc disease, is the most common cause of chronic LBP in adults (DePalma et al., 2011). Non-neutral trunk postures may contribute to this degeneration as they result in static loading of the soft tissues and an accumulation of metabolites that accelerates degeneration leading to disc herniation (Lyons et al., 2002;

Pelham et al., 2005; Pope et al., 2002). However, the precise mechanisms of degeneration remain largely unspecified (Leboeuf-Yde et al., 1997; Peng et al., 2005; Marras, 2012).

Shoulder pain may be attributed to various pathologies originating from the neck, glenohumeral joint, acromioclavicular joint, rotator cuff, and other soft tissues (Walker, 2014; Murphy and Carr, 2010). Abnormalities of the coracoacromial arch and changes in shoulder kinematics have also been theorized to contribute to the etiology of MSDs such as shoulder impingement syndrome (Ludewig and Cook, 2000; Zuckerman et al., 1992). Working with the arms elevated >60º may be hazardous as it is where the space between the humeral head and the acromion begins to narrow such that mechanical pressure on the supraspinatus tendon is greatest (Levitz and Iannotti, 1995). The increased pressure may lead to degenerative changes of the tendons of the rotator cuff, predisposing workers to tears (Armstrong et al., 1993; Nirschl, 1988; Svendsen et al., 2004a, 2004b). Similar to the low back, however, shoulder pain likely has a multifactorial underlying pathology and contributing factors remain difficult to determine definitely (Meislin et al., 2005).

The high prevalence of MSDs among the general population implies that many factors not amenable to prevention, such as age and gender, likely play a significant role in their development. However, quantifying the specific role of these non-amenable factors is difficult to accomplish. Furthermore, the fact that they exist does not rule out the possibility that specific risk factors such as exposure to non-neutral working postures may further increase MSD risk within certain sub-groups (Punnett, 2014).

Methodological limitations of certain study designs and inadequate control of potential confounds may also contribute to the heterogeneity of reported associations. For example, cross-sectional studies are often susceptible to cause-effect reversal bias, a form of error in which the temporal relationship between exposure and health outcome cannot be established definitively. This may lead to results that are not representative of the true causal association. Such inconsistencies in the epidemiological evidence associating non-neutral working postures with MSDs have led many investigators to the conclusion that research into more accurate and precise exposure assessment instruments and strategies is necessary (Garg and Kapellusch, 2009; Marras et al., 2009).

Assessing Exposure to Non-neutral Postures in the Workplace

Three general categories of methods have been used to assess occupational exposure to non-neutral postures. These include self-report, observational, and direct measurement methods (Burdorf and van der Beek, 1999; David, 2005; Dempsey, McGorry, and Maynard, 2005; Li and Buckle, 1999; Winkel and Mathiassen, 1994; Teschke et al., 2009; Vieira and Kumar, 2004). Self-report methods use information provided directly from study participants about their feelings, attitudes, or perceived behaviors. These methods often include interviews, questionnaires, or ratings of perceived exertion (e.g. Borg CR-10 scale) and may be advantageous in comparison to observational and direct measurement techniques because they are simple to distribute, low in cost, are applicable to a wide range of situations, and may provide a useful estimate of some work exposures for population studies (Dale et al., 2010). Self-report studies have validity issues, however, as participants may exaggerate or under-report information in an effort to make their working situation seem better or worse to an investigator (Barriera-Viruet et al., 2006; David, 2005; Hansson et al., 2001b; Teschke et al., 2009). They also often lack precision and accuracy that may result in exposure misclassification (Burdorf and van der Beek, 1999).

Observational methods involve an examiner observing and analyzing the behaviors and actions of a participant as they occur in a natural setting or afterwards using computerized software. These methods are designed to provide quasi-objective estimates of exposure that are not subject to the potentially inaccurate and imprecise information that is often associated with self-report. Examples of widely used observational methods for exposure estimation to non-neutral postures include the Ovako Working position Analyzing System (OWAS) (Karhu et al., 1977), the Rapid Entire

Body Assessment (REBA) (Hignett and McAtamney, 2000), Multimedia Video Task Analysis (MVTA) (Yen and Radwin, 1997), and the Strain Index (Moore and Garg, 1995). Observational methods are relatively easy to perform and have been shown to provide ecologically valid and reliable information (Dartt et al., 2009; Kazmierczak et al., 2006; Li and Buckle, 1999; Takala et al., 2010). However, these methods can be resource intensive, may suffer from observer errors, and often lack criteria for determining the optimum number of observations for analysis of repetitive tasks (Genaidy et al., 1994; Rezagholi et al., 2012).

Direct measurement of a physical risk factor occurs when an exposure is directly measured through the use of a hand-held or electronic measurement tool. Direct measurement provides the most precise estimates and informational content for estimation of exposure to physical risk factors associated with MSDs (Burdorf and van der Beek, 1999; David, 2005; Li and Buckle, 1999; Winkel and Mathiassen, 1994; Teschke et al., 2009; Trask et al., 2007; Vieira and Kumar, 2004). However, these methods are often accompanied with high equipment costs and time demands for calibration and analysis. Field-based direct measures of physical exposures have also been limited in challenging work settings and generate a large amount of raw data that must be appropriately reduced and synthesized to produce relevant summary measures. Common approaches for directly measuring human trunk and shoulder posture include electrogoniometry, optical motion capture (OMC), and body-mounted electromechanical sensors.

Electrogoniometry

Electrogoniometers are devices that use transducers such as potentiometers and strain gauges to quantify an angle and changes of angles between body segments connected by a joint (Li and Buckle, 1999; NIOSH, 2014; Vieira and Kumar, 2004). Biometrics electrogoniometers (Biometrics Ltd., Ladysmith, VA), for example, are commonly used in ergonomics research to measure single axis joints (e.g., the knee or elbow) and consist of two small blocks attached with a flexible wire. A strain gauge allows for conversion of angular displacement between the blocks into a measurable, variable voltage output. By affixing one block to the distal end of a joint and the second block to the proximal end of a joint, the electrogoniometer can be used to measure the angular displacement that occurs between that joint during motion.

The Lumbar Motion Monitor (LMM) is a triaxial electrogoniometer that was developed to measure three-dimensional human trunk motion during occupational activities (Garg and Kapellusch, 2009; Li and Buckle, 1999; Marras et al., 1990, 1992, 1993, 1995). The LMM is an instrumented exoskeleton of the spine that may be attached with a harness to the thorax and pelvis of a worker and allows for continuous measurement of angular displacement in the three primary trunk motion planes (flexion/extension, lateral bending, and axial rotation). Numerical differentiation of the angular displacement time series is then used to obtain estimates of trunk angular velocities and angular accelerations. The LMM has been shown to be accurate and repeatable in controlled experimental settings (Marras et al., 1992; Gill and Callaghan, 1996) and has been used in numerous studies (e.g., Elford et al., 2000; Ferguson et al., 2002; Gallagher et al., 2002; Marras et al., 1999, 2004; Paquet et al., 2001).

While simple in their construction and use, electrogoniometers such as the LMM may restrict natural movement, causing participants to modify their natural motion patterns (Marras et al., 2006). Additionally, the devices may suffer from a phenomenon known as “cross-talk” that occurs when there is rotation between the two end blocks of the electrogoniometer because of soft tissue motion (Buchholz and Wellman, 1997; Hansson et al., 2004). Thus, electrogoniometers have been suggested as a poor measurement tool for analyzing joints that have the ability to move in multiple degrees of freedom such as the back and shoulder (Vieira and Kumar, 2004).

Optical Motion Capture

Video-based OMC technology is considered the gold standard of human motion analysis (Cuesta-Vargas et al., 2010). With many applications including medical diagnostics, detection and surveillance tracking, and activity monitoring (Aggarwal and Cai, 1997; Andriacchi and Alexander, 2000; Wang, Hu, and Tan, 2003), OMC systems record and translate human movement into a digital model through the use of active (light emitting) or reflective markers that are secured to the body and aligned with bony landmarks. Infrared cameras collect the markers' position information and dynamic motion information is calculated using multiple 2D images.

OMC systems have been observed to be accurate and repeatable (Miller et al., 2002) and are considered advantageous in comparison to other motion analysis systems as they operate through the use of passive sensors. That is, the markers used to track human movement are only reflective or light emitting surfaces that can easily be attached to the body without the need for wires or a direct connection to a data logger. Also, such systems only require three markers to define three-dimensional velocity and acceleration of each body segment (Rahmatalla et al., 2006; Verstraete and Soutas-Little, 1990).

While human motion has been monitored proficiently using OMC technologies in the laboratory setting, OMC systems do suffer from a few shortcomings. First, OMC systems are subject to occlusion, a line of sight issue that occurs when the reflective markers do not appear in enough of the camera shots due to blockages between the marker and the cameras by the subject's body or other objects in the scene. Occlusion can be problematic as it requires the use of interpolation to estimate the position of markers and the subject of interest. Interference from other light sources or reflections can also cause false readings known as "ghost markers" that may cause errors in position estimation similar to interpolation errors caused by occlusion. OMC systems are also expensive, require extensive training, and are limited to a controlled laboratory setting with special hardware to obtain and process the data. These shortcomings prevent widespread adoption of OMC as a viable method for quantifying occupational exposure to non-neutral postures.

Body-mounted Electromechanical Inertial Sensors

Inertial sensors are dead-reckoning devices that provide position, velocity, acceleration, and/or orientation information of an object through direct measurements (Altun et al., 2010). Traditionally used in aircrafts, ships, and land vehicles to provide a reference for attitude and heading information, inertial sensors provide investigators a convenient method for capturing human posture and movement information. In ergonomics research, inertial sensors commonly refer to body-mounted accelerometers, gyroscopes, and magnetometers (Gouwanda and Senanayake, 2008).

An accelerometer (or inclinometer) is an electromechanical device that measures the physical acceleration experienced by an object. Single and multi-axis accelerometers exist and detect the magnitude and direction of acceleration as a vector quantity. Forces that cause an accelerometer to register may be static, such as the constant force of gravity, or they may be dynamic, such as vibration or coordinate acceleration. If the line of gravity is used as a reference, two of the three degrees of freedom may be used for measuring angles of slope (tilt) or elevation of an object (Hansson et al., 2001a).

The light-weight, miniature size, and increasing affordability of accelerometers make them a practical method for capturing estimates of trunk and shoulder posture for field-based research. Trunk and shoulder posture estimates have been reported in a number of studies (Douphrate et al., 2012; Fethke et al., 2011; Forsman et al., 2002; Hansson et al., 2006, 2010; Jonker et al., 2009; Paquet, Punnett, and Buchholz, 2001; Svendsen et al., 2004a; Van Driel et al. 2013; Wong, Lee, and Yeung 2009). Accelerometers do, however, suffer from several limitations. First, rotations about the line of gravity cannot be assessed by an accelerometer. This means that arm movements in the sagittal plane (flexion/extension) cannot be separated from movements in the coronal plane (abduction/adduction) and trunk motions about the transverse plane (axial rotation) cannot be assessed through the use of an accelerometer alone. Additionally, the accuracy of accelerometer measurements depends greatly on the characteristics of the motion sampled as they are sensitive to linear accelerations and may be influenced by body characteristics (Amasay et al. 2009; Amasay et al., 2013;Bernmark and Wiktorin, 2002; Giansanti, 2006; Hansson et al. 2001a; Henriksen et al., 2007; Van Driel et al., 2013).

Other sensors capable of providing orientation measurements include gyroscopes and magnetometers (Luinge et al., 2007). A gyroscope is a device that measures angular velocity. Fundamentally, a mechanical gyroscope is a gimbaled wheel or disk whose axle is free to take any orientation and can be set to rotate in any plane, independent of forces tending to change the position of the axis. Thus, unlike accelerometers, gyroscope measurements are not subject to acceleration. When velocity measurements obtained via a gyroscope are integrated, changes in the orientation of the gyroscope can be estimated. Estimation of orientation change with use of a gyroscope, however, is prone to large integration errors, often restricting the time of accurate measurements to less than one minute (Luinge and Veltink, 2005). Gyroscopes are therefore rarely used alone to make estimates of human posture.

A magnetometer is a device used to measure the direction and strength of a magnetic field at a point in space. Similar to a gyroscope, the data obtained via a magnetometer is of little use for ergonomic assessment of human posture without additional information. Measurements of the Earth’s magnetic field, the primary source of data for a magnetometer, are subject to substantial error in the vicinity of ferromagnetic metals and electronics equipment and have been observed to be inaccurate over extended periods of time (de Vries et al., 2010; Luinge et al., 2007; Roetenberg et al., 2005).

Inertial Measurement Units and Ergonomic Applications

An inertial measurement unit (IMU) is a solid-state device that measures and reports an object's orientation and motion characteristics using information collected from multiple body-mounted inertial sensors (i.e., accelerometers, gyroscopes, and/or magnetometers). Specifically, orientation estimates are calculated by fusing the information obtained from some or all of the individual inertial sensors included in the IMU using processing algorithms such as a complementary weighting algorithm or a Kalman filter (Bachmann et al., 1999; Gallagher, Matsuoka, and Ang, 2004; Higgins, 1975; Kalman, 1960; Ligorio and Sabatini, 2013; Luinge and Veltink, 2005; Sabatini, 2006, 2011; Wagenaar et al., 2011; Yun and Bachmann, 2006). Theoretically, IMUs are advantageous to individual inertial sensors as the fusion of information from multiple sources allows for compensation of each individual sensor's limitations. The accelerometer and magnetometer components of an IMU, for example, may be used to help correct for the drift that has been observed to effect an IMU's gyroscope (Bachman et al., 2007; Favre et al., 2006; Luinge and Veltink, 2005; Wong and Wong, 2008; Zhou et al., 2006; Zhu and Zhou, 2004).

Manufacturers of commercially available IMU systems commonly report laboratory-based orientation accuracies of ≤ 2 degrees root-mean-square deviation (RMSD) under dynamic conditions (Xsens Technologies, Enschede, The Netherlands). IMUs have been observed to reliably and consistently estimate joint kinematics of the upper arm and shoulder (Cutti et al., 2008; de Vries et al., 2010; El-Gohary and McNames, 2012; Godwin et al., 2009; Zhou et al., 2007, 2008, 2010), the cervical spine (Duc et al., 2014; Theobald et al., 2012; Jasiewicz et al., 2007), the lower extremity (O'Donovan et al., 2009; Favre et al., 2008; Ferrari et al., 2009; Fong and Chan, 2010; Picerno et al., 2008), the trunk (Giansanti et al., 2007; Goodvin et al., 2006; Kim and Nussbaum, 2013; Lee et al., 2003), and the whole body (Brodie et al., 2008b) in comparison to laboratory-based human motion analysis techniques such as OMC systems

(Cuesta-Vargas et al., 2010). They have been used to detect gait events and parameters (Aminian et al., 2002; Catalfamo et al., 2010; Greene et al., 2010; Lee and Park, 2011), to recognize human activities (Altun et al., 2010), to measure postural stability (Frames et al., 2013), and to analyze the combined effect of acceleration and posture (Dickey et al., 2013).

While many of the aforementioned studies have addressed the basic concern of evaluating the accuracy of measurements obtained with IMUs in comparison to OMC systems, many limitations of IMU technology remain that deter investigators from using them in epidemiological field studies where associations between exposure to non-neutral posture and MSDs may be explored. For instance, investigations of the performance of IMU systems in field settings and against other field-capable technologies including individual sensor components of the IMU are still needed to better establish the accuracy and reliability of IMUs as proficient direct measurement devices for field-based applications. IMUs developed solely in the controlled setting of a laboratory are not guaranteed to work in the diverse and complex work environments that often accompany real-world data collection. Studies examining the performance of IMUs in a field environment and against other field-capable reference instruments must be conducted in order to establish IMUs as effective direct measurement tools and to prevent researchers from underestimating the time, personnel, and monetary resources required to obtain the data necessary for answering their research questions (Trask et al., 2007).

Another limitation of current IMU technology is a lack of research comparing different sensor configurations (wear locations) and processing methods (fusion algorithms) that may be used to generate orientation estimates from IMUs. While investigators have developed and successfully used fusion algorithms such as the complementary weighting algorithm and the Kalman filter to obtain and improve direct measurements of human motion (Favre et al., 2006; Luinge et al., 2005; Luinge and Veltink, 2007), the information used for these studies has generally been collected in a laboratory environment where interactions with ferromagnetic materials and complex work tasks are controlled. Furthermore, many filtering techniques such as the Kalman filter are computationally demanding and may be considered too complex for many investigators to comfortably apply and understand. Further research evaluating the accuracy and repeatability of several different sensor configurations and processing methods on the estimates of exposure to non-neutral trunk and shoulder postures obtained with the IMU system are needed.

Summary and Specific Aims

Occupational exposure to non-neutral working postures has been associated with the development of MSDs of the low back and shoulder in many populations. While direct measurement methods such as electrogoniometery, OMC, and individual body-mounted electromechanical sensors have been used to provide estimates of work-related exposure to non-neutral postures in past epidemiological studies, the limited accuracy and applicability of such measurement methods for field-based applications has restricted the ability to estimate true associations between exposure to non-neutral postures and MSDs. A growing emphasis has therefore been placed on the development of instrumentation devices better suited for field-based occupational exposure assessment.

IMUs have been observed to reliably and consistently estimate joint kinematics in comparison to laboratory-based OMC systems over relatively short time periods. Despite their strong performance in the laboratory setting, IMU technologies have rarely been used in field-based studies characterizing the association between exposure to non-neutral working postures and MSDs. Limited research comparing exposure information obtained with IMUs to exposure information obtained with other field-capable direct measurement exposure assessment methods and a deficiency of knowledge regarding different sensor fusion algorithms and processing methods on estimates of exposure contribute to the lack of IMU use in field studies.

This thesis was designed to address these issues and expand upon the current scientific literature regarding IMU sensors as a direct measurement tool for assessing exposure to non-neutral postures among workers in real work environments. Three specific aims were developed in support of this goal:

SA #1: Evaluate the accuracy and repeatability of a commercially available IMU system for quantifying exposures to non-neutral trunk and shoulder postures for use in field-based occupational studies.

SA #2: Explore the effect of several different sensor configurations and processing methods on the estimates of exposure to non-neutral trunk and shoulder postures obtained with the IMU system.

SA #3: Apply the IMU system in a field-based occupational study to estimate exposures to non-neutral trunk and shoulder postures to demonstrate the utility of IMUs as direct measurement instruments.

The remainder of the thesis is organized as follows. Chapter II presents the results of a manual material handling study that was performed to compare estimates of thoracolumbar trunk posture obtained with a commercially available IMU system to a field-capable system, the LMM. A second objective of the study was to explore the effect of alternative sensor configurations and processing methods on the agreement between LMM and IMU-based estimates of thoracolumbar trunk motion. Chapter III presents the results of a study performed to evaluate the accuracy and repeatability of estimates of trunk angular displacement and upper arm elevation obtained with the IMU system examined in Chapter II over the course of an eight-hour work shift in both a laboratory and field-based setting. Chapter IV presents the results of a randomized, repeated measures intervention that demonstrates the utility of the IMU system examined in Chapters II and III as a useful direct measurement tool for comparing “ergonomic” and conventional examination equipment commonly used by Ophthalmologists. Finally, Chapter V summarizes the major findings, discusses their practical implications, and provides suggestions for future research.

CHAPTER II

A COMPARISON OF INSTRUMENTATION METHODS TO ESTIMATE THORACOLUMBAR MOTION IN FIELD-BASED OCCUPATIONAL STUDIES

Introduction

Low back pain (LBP) is a common work-related musculoskeletal disorder (MSD) with an estimated 1-month prevalence of 23.2% and lifetime prevalence estimates ranging as high as 84% (Hoy et al., 2012; Walker, 2000). Occupational exposure to non-neutral trunk postures and manual material handling (MMH) activities may be associated with LBP (Coenen et al., 2013; da Costa and Vieira, 2010; Manchikanti, 2000; Hoogendoorn et al., 2000; Kerr et al., 2001; Vieira and Kumar, 2004; van Oostrom et al., 2012). Evidence of these associations, however, is inconsistent (Roffey et al., 2010; Wai et al., 2010a, 2010b). In part, characterization of associations between non-neutral trunk postures and LBP has been limited by use of easily administered but imprecise and potentially biased self-report or observation-based exposure assessment methods (Burdorf and van der Beek, 1999; David, 2005; Li and Buckle, 1999; Vieira and Kumar, 2004).

Common approaches for directly measuring thoracolumbar trunk motion in a field setting include electrogoniometry and body-mounted electromechanical sensors (David, 2005; Li and Buckle, 1999; Vieira and Kumar, 2004). The Lumbar Motion Monitor (LMM) is a field-capable, triaxial electrogoniometer used to directly measure kinematics of the thoracolumbar spine (Marras et al., 1992; Marras and Granata, 1995; Gill and Callaghan, 1996). The LMM is secured to the trunk of a worker using chest and pelvic harnesses and measures thoracolumbar angular displacement of the trunk relative to the pelvis in the three primary motion planes. With software, numerical differentiation of the angular displacement measurements is then used to obtain estimates of trunk angular velocities and angular accelerations in the three motion planes. Although the LMM has been used in numerous studies (e.g., Ferguson et al., 2002; Gallagher et al., 2002; Marras et al., 2004; Marras et al., 1999), its bulky size and limited range (i.e., through direct cable connection to a computer or through telemetry) make it impractical for prolonged field-based exposure assessments recommended to obtain stable and representative estimates of trunk motion during non-routinized work activities (e.g., construction and agriculture) (Trask et al., 2007).

Accelerometers have been used frequently in field-based research to obtain direct measurements of trunk motion over extended time periods (e.g., Fethke et al., 2011; Koehoorn, 2010; Paquet et al., 2001; Teschke et al., 2009; Van Driel et al. 2013; Wong et al., 2009). Trunk motion estimates have been reported using a variety of sensor configurations (e.g., dual axis or triaxial) and sensor placement strategies (e.g., one sensor placed on the anterior torso as in Fethke et al. [2011] vs. one sensor on the posterior torso as in Wong et al. [2009] vs. one sensor on the anterior torso combined with one sensor on the posterior pelvis as in Koehoorn [2010]). However, axial rotations in the transverse plane cannot be assessed through the use of an accelerometer alone and the accuracy of accelerometer-based estimates in the flexion/extension and lateral bending motion planes depends on the characteristics of the motion (static, quasi-static, or complex dynamic) (Amasay et al., 2009; Brodie et al., 2008a; Godwin et al., 2009; Hansson et al., 2001a).

Inertial measurement units (IMUs) have recently emerged as a potential alternative to accelerometers for measurement of human trunk motion in occupational settings. An IMU is a small and portable device that permits estimation of the spatial orientation of an object by combining the outputs of multiple electromechanical sensors (accelerometers, gyroscopes, and/or magnetometers) through recursive sensor fusion algorithms such as a Kalman filter or complementary weighting algorithm (Bachmann et al., 1999; Gallagher et al., 2004; Higgins, 1975; Kalman, 1960; Ligorio and Sabatini, 2013; Luinge and Veltink, 2005; Sabatini, 2006, 2011; Wagenaar et al., 2011; Yun and

Bachmann, 2006). Theoretically, using sensor fusion algorithms for motion measurement can help overcome the limitations of each individual sensor component. For example, gyroscope measurements can be used to compensate for limitations of the accelerometer to more accurately measure motion in the flexion/extension and lateral bending planes under dynamic conditions and magnetometers can provide orientation information necessary to make estimates of trunk motion in the axial rotation plane. Raw sensor streams from the individual sensor components may also be extracted for singular analysis.

Despite their unique capabilities and promise, few studies have used IMUs to directly measure thoracolumbar trunk motion in the field. One potential explanation for their limited use may be a lack of justification in comparison to more widely known methods such as accelerometers or electrogoniometer systems such as the LMM. While many studies have examined the accuracy of IMU systems in comparison to optical motion capture (OMC) systems (Cuesta-Vargas et al., 2010) and/or have evaluated corrective factors for accelerometers (e.g., Van Driel et al., 2013), the potential benefit of using IMUs to estimate thoracolumbar motion in comparison to other field-capable systems remains unclear. For example, estimates of trunk motion can be made using information obtained from an IMU's accelerometer alone, from an IMU's accelerometer and gyroscope, or from the full complement of IMU sensors (i.e., accelerometers, gyroscopes, and magnetometers). Exploration of the different sensor configurations and processing methods possible with an IMU system will provide information about the potential advantages of IMU use in comparison to simpler options.

The objectives of this study were, therefore, to (1) compare estimates of thoracolumbar trunk motion obtained with a commercially available IMU system with estimates of thoracolumbar trunk motion obtained with a field-capable reference system, the LMM, and to (2) explore the effect of alternative sensor configurations and processing methods on the agreement between LMM and IMU-based estimates of trunk motion during a simulated MMH task with both systems deployed simultaneously.

Methods

Participants

A convenience sample of 36 healthy, male participants (mean age=24.9 years, SD=4.5) was recruited from the University of Iowa community. Potential participants were excluded for any self-reported 1) physician-diagnosed MSDs of the back in the past six or fewer months, 2) orthopedic surgery of the back, 3) back pain in the past two weeks, or 4) chronic neurodegenerative disease (e.g., Parkinson's disease). All study procedures were approved by the University of Iowa Institutional Review Board and written informed consent was obtained prior to participation.

Experimental Design

Participants completed a simulated MMH task in a laboratory setting. The MMH task required participants to manually move 4.5 kg plastic crates (42 × 35 × 27 cm) from a waist-high material feeder (Point A in Figure 1, as depicted from above) to one of six potential unloading areas (Point B in Figure 1). Two handholds were molded into each crate and used by workers for manual grasping. The six potential unloading areas varied across two factors: the unloading height (adjusted to each participant to be approximately waist height or knee height) and the total magnitude of axial rotation needed to move a crate from the material feeder to the unloading area (90°, 135°, or 180°). The pace of the task was set to either 6 lifts/min or 3 lifts/min. Block randomization was used to assign each participant to one of the 12 task conditions (2 unloading heights × 3 axial rotation magnitudes × 2 work paces; 3 participants per condition). The modest crate weight and work pace levels were selected to ensure that the recommended weight limit of the NIOSH Lifting Equation was not exceeded when considering all combinations of the unloading height, amount of axial rotation, and work pace parameters (Waters et al., 1993).

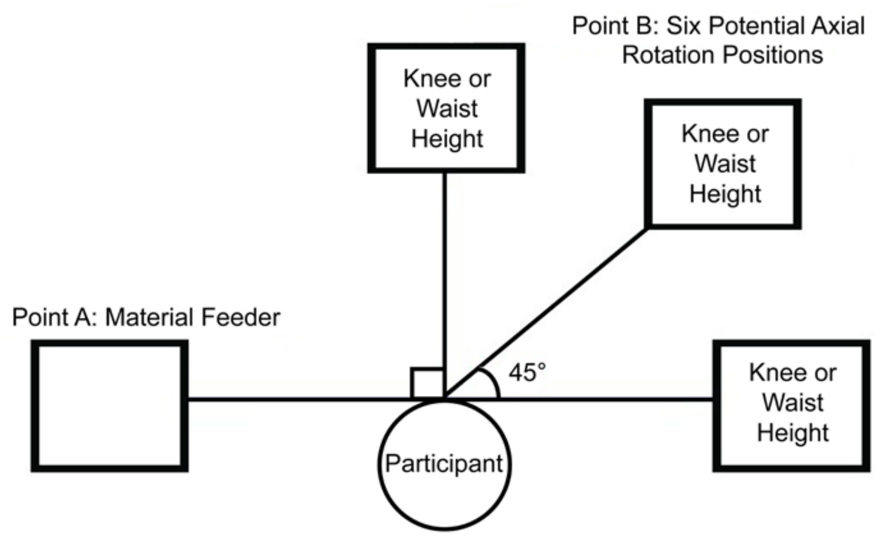

Figure 1. Simulated manual material handling task positions.

A custom LabVIEW program was used to control the simulated MMH task. The program produced an auditory tone at the assigned work pace to signal the participant when to move each crate. Data collection began with the participant standing in a neutral upright stance and the arms hanging relaxed and the feet hip-width apart. At each auditory tone, the participant would, 1) rotate left to the material feeder location and grasp the crate’s handles using two hands, 2) rotate right and set the crate on the assigned unloading location, and then 3) return to the neutral standing position to wait for the next auditory tone. Participants were instructed to keep their feet stationary while performing the lifts and to use the crate handholds. The material feeder and unloading locations were set close to the body (within arm’s reach) for all participants. No other instructions were given. The task was performed for 10 minutes, resulting in 30 or 60 lifting cycles, depending on the assigned work pace.

Instrumentation and Data Processing

Angular displacements of the thoracolumbar region of the trunk in the flexion/extension, lateral bending, and axial rotation motion planes were estimated using two commercially available instrumentation systems: the ACUPATH™ Industrial Lumbar Motion Monitor™ (Biomec Inc., Cleveland, OH, USA) and the I2M Motion Tracking System (series SXT IMUs, Nexgen Ergonomics, Inc., Pointe Claire, Quebec, CAN). For each participant, one IMU sensor was secured to the anterior torso at the sternal notch and a second IMU sensor was secured to the posterior pelvis at the L5/S1 vertebrae. Standard procedures were used to outfit participants with the LMM as in previous studies (Marras et al., 1993). The LMM was calibrated prior to fitting by using procedures described in the LMM manual. Data streams obtained from the LMM included angular displacement (in degrees) of the trunk in the flexion/extension and lateral bending motion planes. The LMM was connected to a computer using a communications cable and the data streams were sampled at 60 Hz using manufacturer-supplied software (Ballet 2.01, Biomec Inc., Cleveland, OH).

The small size of the IMU sensors (48.5 × 36 × 12 mm) allowed them to be worn simultaneously with the LMM. The IMU sensors were calibrated after the LMM was fit using an “I-pose” calibration posture in which each participant stood in a neutral trunk position with hands relaxed to the sides and the feet hip-width apart as if forming an “I”. Calibration quality was visually inspected before beginning the MMH task. Data streams obtained from each IMU sensor included acceleration (triaxial, +/- 6 g), angular velocity (triaxial, +/- 2000° s^{-1}), magnetic field strength (triaxial, +/- 6 Gauss), and local sensor spatial orientation in the form of quaternions derived from a manufacturer-provided Kalman filter. The IMU data streams were sampled wirelessly at 128 Hz using manufacturer-supplied software (HM Analyzer, Nexgen Ergonomics, Pointe Claire, Quebec, CAN). An event marker was used during data collection to facilitate synchronization of the LMM and IMU data during post-processing. A custom LabVIEW program (version 2012, National Instruments, Austin TX) down sampled the IMU estimates of trunk motion from 128 Hz to 60 Hz using linear interpolation and exported the data for later analysis.

Five IMU processing methods were used to obtain estimates of thoracolumbar angular displacement in the flexion/extension and lateral bending motion planes. The five methods were: 1) a low passed (zero-phase, 2nd order Butterworth; 3 Hz cutoff frequency) accelerometer-based estimate from the IMU secured to the sternum only (Accel-1); 2) a complementary weighting algorithm-based estimate incorporating accelerometer and gyroscope measurements from the IMU secured to the sternum only (Comp-1); 3) a low passed (zero-phase, 2nd order Butterworth; 3 Hz cutoff frequency) accelerometer-based estimate calculated as the difference of the estimates provided from the IMUs secured to the sternum and L5/S1 body segments (Accel-2); 4) a complementary weighting algorithm-based estimate calculated as the difference of complementary-based estimates from the IMUs secured to the sternum and L5/S1 body segments (Comp-2); and 5) a manufacturer provided Kalman-based estimate which incorporated raw acceleration, angular velocity, and magnetic field strength information from the IMU located on the sternum and L5/S1 body segments (HM Analyzer). The manufacturer provided Kalman-based estimate was also used to provide estimates of thoracolumbar angular displacement in the axial rotation motion plane.

Accelerometer-based angular displacement estimates reflected accelerometer inclination angle with respect to the gravity vector and were calculated as the arctangent of the acceleration reading pointing away from the sternum (i.e., the z-axis of the SXT IMU) and the acceleration reading corresponding to the gravity vector (i.e., the x-axis of the SXT IMU and not the norm of gravity). For example, to estimate angular displacement in the flexion/extension motion plane, the accelerometer-based inclination angle estimates from the IMU secured to the sternum only (Accel-1) were calculated as $\tan^{-1}(A_z / A_x)$. Accelerometer-based angular displacement estimates were calculated in this manner so that they could be paired with gyroscope measurements in the corresponding axis of rotation.

The custom complementary weighting algorithm was developed in MATLAB (r2013b, The MathWorks, Inc., Natick, MA) and used the raw data streams of acceleration and angular velocity to estimate the trunk motion angles from the orientation of the IMU's accelerometer with respect to the gravitational vector and angular velocity information from the IMU's gyroscope. The complementary weighting algorithm adjusted the accelerometer-based inclination angle estimate at each sample time using angular velocity information from the IMU's gyroscope according to Equation (1):

$$\theta_n = (1 - K)\,[\theta_{n-1} + (\omega_n \times dt)] + K(\alpha_n) \qquad \text{(Equation 1)}$$

where θ_n is the complementary inclination angle estimate at the current sample, $\theta_{n-1}$is the complementary inclination angle estimate at the previous sample, ω_n is the angular velocity at the current sample, α_n is the inclination angle at the current sample based solely on the orientation of the accelerometer with respect to gravity, and *dt* is the time between samples. The algorithm's coefficient (K) weighted the relative influence of the angular velocity and the accelerometer-based inclination angle on the resulting complementary inclination angle estimate. Although there are no widely accepted guidelines for selecting the weighting coefficient, a value of 0.01provided a sufficient acceleration reference to compensate for the drift that occurs when a raw gyroscope signal is integrated (Luinge and Veltink, 2005).

The complementary weighting algorithm had a time constant of 0.77 sec, based on the weighting coefficient, the sampling rate (128 Hz), and the IMU gyroscope drift rate (approximately 1° s^{-1}). The inclination angle (α_n) was low pass filtered (zero-phase, 2^{nd} order Butterworth; 3 Hz cutoff frequency) and the angular velocity (ω_n) signals were high pass filtered (zero-phase, 2nd order Butterworth; 0.5 Hz cutoff frequency) prior to computation of complementary inclination angle estimates.

Statistical Analysis

Using the angular displacement waveform obtained from the LMM for each participant, a custom MATLAB program was used to identify the peak (maximum) point of flexion for the flexion/extension motion plane, lateral bending to the right for the lateral bending motion plane, and axially rotating to the left for the axial rotation motion plane of each lifting cycle and the corresponding four seconds before and after each peak. This eight second window encompassed all phases of each lift cycle for all participants (e.g., start of lift, peak flexion, and end of lift). The arithmetic mean of each respective sample estimate from the 30 or 60 cycles comprising the entire 10 minute MMH task was then calculated to form an ensemble average of a lifting cycle lasting eight seconds in length (ensemble averages were generated for each participant separately).

Ensembles averages of the angular displacement waveforms were differentiated to obtain an ensemble average waveform estimate of velocity. The velocity waveform was rectified to represent the absolute value of velocity (indicating either increasing or decreasing speed). The rectified ensemble average waveform of velocity was differentiated to obtain an ensemble average waveform of acceleration (Marras et al., 1995). The minimum, maximum, mean, 10th percentile, 90th percentile, and 99th percentile were then calculated for each ensemble average waveform. In addition, we calculated a sample-to-sample root-mean-square difference (RMSD) of the ensemble average waveforms obtained with each IMU processing method in comparison to the ensemble average waveforms obtained with the LMM. The RMSD for each participant was calculated using Equation (2), where θ was the estimate from an IMU processing method, θ' was the estimate from the LMM, n was the number of samples included in the ensemble waveform, and i was the specific sample of interest.

$$\text{RMSD} = \sqrt{\sum_{i=1}^{n}(\theta_i - \theta_i')^2/n} \qquad \text{(Equation 2)}$$

Pearson correlation analyses were used to quantify the strength of the linear relationships between the estimates of mean angular displacement and angular displacement variation (defined as the difference between the estimates of the 90th and 10th percentile) from the LMM and each measurement method in the flexion/extension, lateral bending, and axial rotation motion planes. Bland and Altman (1986, 1995, 1999, 2010) bias calculations and 95% limits of agreement (LoA) were used to assess agreement between estimates of mean angular displacement for the flexion/extension, lateral bending, and axial rotation motion planes obtained with the LMM and each applicable IMU measurement method.

Results

The LMM and each of the IMU measurement methods produced waveforms of trunk angular displacement with similar characteristics (Figure 2). In general and consistent with our expectations, estimates of mean angular displacement in the flexion/extension motion plane were lower for participants assigned to the waist high unloading areas in comparison to participants assigned to the knee high unloading areas. Moreover, the greatest estimates of mean angular displacement in the flexion/extension motion plane were observed for participants assigned to the knee high unloading areas and the faster work pace.

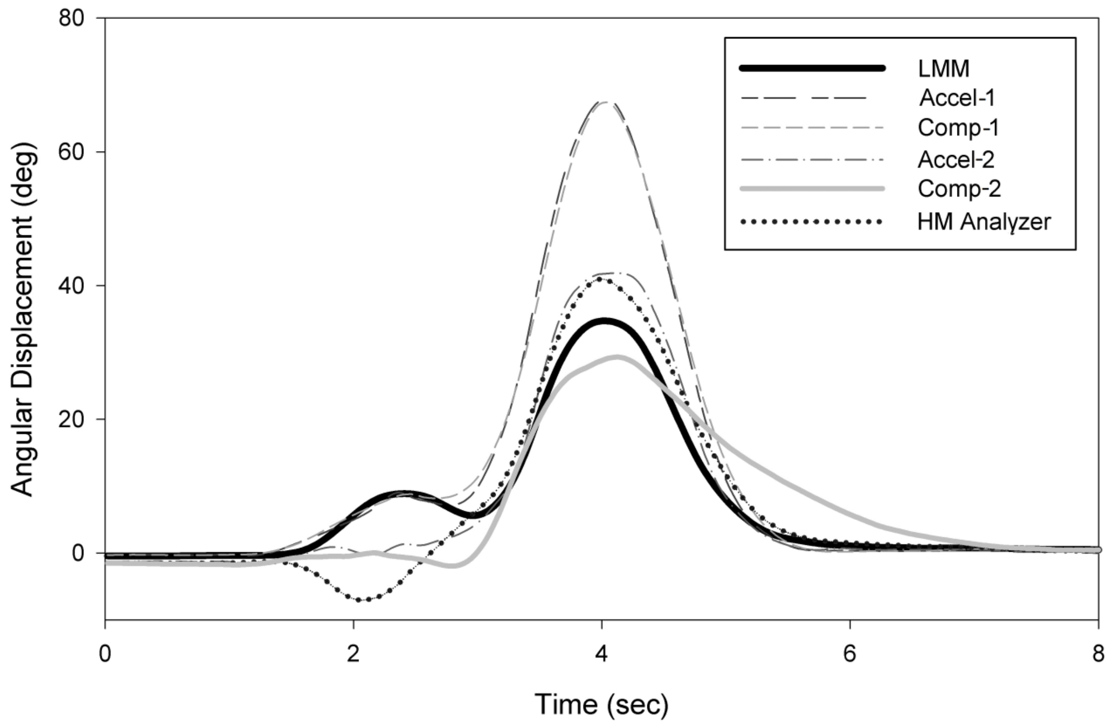

Figure 2. Ensemble average of angular displacement waveform for the LMM and the five IMU measurement methods in the flexion/extension motion plane for one participant.

Distributions of summary measures of trunk angular displacement, velocity, acceleration, and estimates of the RMSD between the LMM and the IMU measurement methods for the flexion/extension, lateral bending, and axial rotation motion planes are provided in Tables 1-7. RMSD estimates were similar across methods obtained using only the IMU secured to the sternum and across methods obtained using both the IMUs secured to the sternum and L5/S1 body segment. In general, the methods that used the IMUs secured to the sternum and L5/S1 body segment were observed to be more comparable to the LMM than methods obtained with IMUs secured to the sternum only. Summary measure estimates such as the mean, 10th percentile, 90th percentile, and 99th percentile angular displacement were the most comparable with the LMM for the complementary weighting algorithm-based method that calculated the estimate of thoracolumbar angular displacement as the difference of complementary-based estimates provided from the IMUs secured to the sternum and L5/S1 body segments (Comp-2). Similarly, estimates of mean, 10^{th} percentile, 90^{th} percentile, and 99^{th} percentile velocity and acceleration obtained with the complementary weighting algorithm-based method that calculated the estimate of thoracolumbar angular displacement as the difference of complementary-based estimates provided from the IMUs secured to the sternum and L5/S1 body segments (Comp-2) were the most comparable to the LMM.

Table 1. Mean (SD) of trunk angular displacement estimates in the flexion/extension motion plane by measurement method for ensemble averages.

Summary measure	LMM	Accel-1	Comp-1	Accel-2	Comp-2	HM Analyzer
Maximum Extension (°)	-1.0 (0.6)	-1.2 (0.8)	-1.2 (0.8)	-2.1 (1.8)	-2.0 (1.7)	-2.4 (2.0)
Mean (°)	3.7 (2.5)	7.6 (5.7)	7.7 (5.7)	5.5 (5.0)	4.9 (4.8)	5.3 (5.2)
Maximum Flexion (°)	17.2 (9.8)	34.9 (27.7)	34.5 (27.0)	26.3 (22.3)	20.3 (17.6)	26.9 (23.6)
10^{th} Percentile (°)	-0.8 (0.5)	-1.0 (0.7)	-0.9 (0.7)	-1.5 (1.5)	-1.6 (2.2)	-1.5 (1.4)
90^{th} Percentile (°)	14.6 (9.4)	30.3 (24.5)	30.2 (23.5)	23.2 (20.4)	17.8 (15.8)	22.8 (21.0)
99^{th} Percentile (°)	17.2 (9.8)	34.8 (27.7)	34.4 (27.0)	26.3 (22.3)	20.3 (17.5)	26.9 (23.6)
Sample-to-Sample RMSD (°)	-- Ref --	8.8 (6.5)	8.9 (6.2)	6.4 (5.2)	6.6 (4.8)	6.7 (5.1)

LMM = Lumbar Motion Monitor; Accel-1= Low passed (zero-phase, 2^{nd} order Butterworth, 3 Hz cutoff) accelerometer-based estimate from the IMU secured to the sternum only; Comp-1 = Complementary weighting algorithm-based estimate using accelerometer and gyroscope measurements from the IMU secured to the sternum only; Accel-2 = Low passed (zero-phase, 2^{nd} order Butterworth, 3 Hz cutoff) accelerometer-based estimate calculated as the difference of the estimates provided from the IMU secured to the sternum and L5/S1 body segments; Comp-2 = Complementary weighting algorithm-based estimate calculated as the difference of complementary-based estimates provided from the IMUs secured to the sternum and L5/S1 body segments; HM Analyzer = Manufacturer provided Kalman-based estimate which incorporated raw acceleration, angular velocity, and magnetic field strength information from the IMU located on the sternum and L5/S1 body segments

Table 2. Mean (SD) of absolute value of velocity estimates in the flexion/extension motion plane by measurement method for ensemble averages.*

Summary measure	LMM	Accel-1	Comp-1	Accel-2	Comp-2	HM Analyzer
Mean (°/sec)	5.1 (2.7)	9.5 (7.0)	9.0 (6.8)	7.7 (5.8)	5.8 (4.7)	7.6 (6.0)
Maximum (°/sec)	24.5 (12.7)	44.1 (33.6)	37.1 (30.1)	38.5 (30.0)	22.6 (17.4)	33.1 (26.4)
10th Percentile (°/sec)	0.1 (0.1)	0.2 (0.1)	0.1 (0.1)	0.2 (0.1)	0.3 (0.4)	0.1 (0.1)
90th Percentile (°/sec)	17.8 (9.8)	32.6 (27.0)	30.0 (25.7)	26.2 (21.8)	14.3 (11.4)	24.8 (21.3)
99th Percentile (°/sec)	24.4 (12.7)	44.0 (33.7)	36.8 (30.1)	38.3 (30.0)	22.1 (17.2)	32.8 (26.5)
Sample-to-Sample RMSD (°/sec)	-- Ref --	10.9 (7.5)	10.1 (6.8)	9.0 (6.1)	7.3 (4.3)	8.7 (6.0)

* Column headers are defined in text and Table 1

Table 3. Mean (SD) of acceleration estimates in the flexion/extension motion plane by measurement method for ensemble averages.*

Summary measure	LMM	Accel-1	Comp-1	Accel-2	Comp-2	HM Analyzer
Maximum Deceleration (°/sec^2)	-72.8 (31.4)	-138.2 (69.8)	-133.2 (92.1)	-146.7 (74.3)	-120.9 (93.3)	-153.5 (104.1)
Maximum Acceleration (°/sec^2)	83.7 (46.0)	141.4 (80.2)	140.8 (95.9)	166.9 (113.0)	125.9 (82.4)	147.1 (88.7)
10th Percentile (°/sec^2)	-29.1 (14.5)	-54.0 (36.1)	-43.1 (30.8)	-52.2 (27.5)	-30.8 (17.7)	-42.3 (25.7)
90th Percentile (°/sec^2)	30.5 (15.5)	57.1 (38.2)	46.2 (33.6)	52.4 (27.1)	33.5 (20.1)	44.3 (28.5)
99th Percentile (°/sec^2)	75.8 (43.2)	123.3 (74.8)	107.1 (73.6)	133.7 (79.8)	87.9 (54.4)	103.0 (63.3)
Sample-to-Sample RMSD (°/sec^2)	-- Ref --	46.4 (23.7)	41.0 (21.7)	47.4 (21.3)	39.6 (18.2)	42.3 (20.3)

* Column headers are defined in text and Table 1

Table 4. Mean (SD) of trunk angular displacement estimates in the lateral bending motion plane by measurement method for ensemble averages.*

Summary measure	LMM	Accel-1	Comp-1	Accel-2	Comp-2	HM Analyzer
Maximum to the Left (°)	-2.2 (1.6)	-3.8 (2.6)	-7.9 (4.8)	-4.9 (2.6)	-3.9 (2.0)	-8.5 (5.9)
Mean (°)	0.5 (0.9)	0.1 (0.8)	0.0 (1.1)	0.1 (0.9)	0.1 (0.7)	0.0 (2.2)
Maximum to the Right (°)	4.8 (4.1)	3.8 (2.9)	8.8 (7.3)	4.8 (3.6)	3.4 (2.3)	8.1 (10.7)
10^{th} Percentile (°)	-1.7 (1.3)	-2.3 (1.7)	-5.4 (3.6)	-2.9 (1.9)	-2.3 (1.5)	-6.4 (4.7)
90^{th} Percentile (°)	3.7 (3.7)	2.6 (2.5)	6.5 (5.6)	3.3 (2.9)	2.4 (2.0)	6.6 (9.7)
99^{th} Percentile (°)	4.7 (4.1)	3.8 (2.9)	8.7 (7.3)	4.7 (3.5)	3.4 (2.3)	8.1 (10.7)
Sample-to-Sample RMSD (°)	-- Ref --	2.1 (1.3)	4.4 (2.5)	2.1 (1.4)	2.2 (1.3)	4.2 (3.5)

* Column headers are defined in text and Table 1

Table 5. Mean (SD) of absolute value of velocity estimates in the lateral bending motion plane by measurement method for ensemble averages.*

Summary measure	LMM	Accel-1	Comp-1	Accel-2	Comp-2	HM Analyzer
Mean (°/sec)	1.9 (1.2)	3.0 (1.7)	5.3 (3.5)	4.0 (1.9)	2.8 (1.3)	4.5 (3.6)
Maximum (°/sec)	7.4 (4.7)	12.8 (7.0)	19.5 (11.5)	18.2 (8.4)	11.5 (5.9)	19.9 (15.4)
10th Percentile (°/sec)	0.1 (0.0)	0.2 (0.1)	0.3 (0.2)	0.3 (0.2)	0.2 (0.1)	0.2 (0.1)
90th Percentile (°/sec)	5.2 (3.7)	7.6 (4.4)	13.9 (9.2)	10.3 (5.3)	7.3 (3.8)	13.2 (11.6)
99th Percentile (°/sec)	7.3 (4.7)	12.3 (7.0)	19.2 (11.6)	17.5 (8.4)	11.2 (5.9)	19.6 (15.5)
Sample-to-Sample RMSD (°/sec)	-- Ref --	3.2 (1.7)	6.4 (4.4)	4.3 (2.1)	3.3 (1.6)	5.5 (4.9)

* Column headers are defined in text and Table 1

Table 6. Mean (SD) of acceleration estimates in the lateral bending motion plane by measurement method for ensemble averages.*

Summary measure	LMM	Accel-1	Comp-1	Accel-2	Comp-2	HM Analyzer
Maximum Deceleration (°/sec^2)	-38.7 (10.3)	-94.3 (44.3)	-86.8 (35.9)	-150.3 (74.4)	-77.4 (47.3)	-100.6 (72.2)
Maximum Acceleration (°/sec^2)	33.6 (14.8)	95.2 (46.6)	94.1 (37.3)	171.8 (99.5)	88.4 (57.1)	126.2 (135.3)
10th Percentile (°/sec^2)	-10.8 (4.2)	-32.4 (13.5)	-29.2 (13.4)	-50.1 (19.0)	-23.7 (8.9)	-30.7 (16.1)
90th Percentile (°/sec^2)	11.0 (5.0)	33.3 (15.1)	29.6 (13.4)	49.2 (18.8)	22.9 (8.1)	30.6 (17.1)
99th Percentile (°/sec^2)	26.3 (13.3)	73.3 (35.0)	69.0 (28.6)	113.0 (47.2)	56.9 (32.5)	75.4 (45.6)
Sample-to-Sample RMSD (°/sec^2)	-- Ref --	28.8 (11.3)	28.1 (9.8)	43.2 (15.8)	23.5 (8.3)	28.0 (13.4)

* Column headers are defined in text and Table 1

Table 7. Mean (SD) of trunk motion estimates in the axial rotation motion plane by measurement method for ensemble averages.*

Summary measure	LMM	HM Analyzer
Angular Displacement		
Maximum to the Left (°)	-13.0 (6.3)	-22.6 (8.1)
Mean (°)	-1.5 (1.1)	-1.3 (2.5)
Maximum to the Right (°)	3.5 (5.2)	11.8 (10.3)
10^{th} Percentile (°)	-10.9 (5.5)	-16.3 (6.6)
90^{th} Percentile (°)	3.2 (4.9)	10.1 (9.0)
99^{th} Percentile (°)	3.4 (5.2)	11.8 (10.2)
Sample-to-Sample RMSD (°)	-- Ref --	6.2 (2.6)
Absolute Value of Velocity		
Mean (°/sec)	4.2 (2.6)	8.8 (3.8)
Maximum (°/sec)	29.8 (18.0)	38.1 (15.8)
10^{th} Percentile (°/sec)	0.0 (0.0)	0.1 (0.1)
90^{th} Percentile (°/sec)	0.8 (0.7)	3.9 (3.1)
99^{th} Percentile (°/sec)	29.8 (17.9)	37.9 (15.9)
Sample-to-Sample RMSD (°/sec)	-- Ref --	9.7 (4.0)
Acceleration		
Maximum Deceleration (°/sec^2)	-83.0 (49.2)	-156.3 (98.3)
Maximum Acceleration (°/sec^2)	164.8 (109.4)	158.4 (72.9)
10^{th} Percentile (°/sec^2)	-27.5 (15.8)	-45.2 (16.8)
90^{th} Percentile (°/sec^2)	20.2 (16.3)	46.7 (18.3)
99^{th} Percentile (°/sec^2)	141.8 (95.9)	113.7 (47.8)
Sample-to-Sample RMSD (°/sec^2)	-- Ref --	45.4 (19.4)

* Column headers are defined in text and Table 1.

Pearson correlation coefficients and Bland and Altman evaluations used to assess agreement between the estimates of mean angular displacement and angular displacement variation between the LMM and each measurement method in the flexion/extension and lateral bending motion planes further suggest that the complementary-based estimates provided from the IMUs secured to the sternum and L5/S1 body segments (Comp-2) generally had the greatest agreement with LMM measures than the other measurement methods (Table 8). Overall, stronger correlation coefficients were observed for estimates of mean angular displacement in the flexion/extension motion plane than estimates of mean angular displacement in the lateral bending motion plane.

Table 8. Pearson correlation coefficients (*r)* of mean angular displacement and angular displacement variation (90^{th} – 10^{th} percentile) and Bland Altman bias and limits of agreement of mean angular displacement in the three trunk motion planes.

Summary Measure	Accel-1	Comp-1	Accel-2	Comp-2	HM Analyzer
Flexion/Extension					
Mean Angular Displacement (*r*)	0.87	0.87	0.84	0.83	0.80
90^{th} - 10^{th} Percentile (*r*)	0.87	0.86	0.87	0.85	0.82
Mean Bias (°)	3.91	3.99	1.82	1.23	1.60
Lower Limit of Agreement (°)	-3.40	-3.27	-4.50	-4.72	-5.37
Upper Limit of Agreement (°)	11.22	11.26	8.14	7.17	8.56
Upper – Lower (°)	14.62	14.53	12.64	11.89	13.93
Lateral Bending					
Mean Angular Displacement (*r*)	0.23	0.08	0.37	0.38	0.42
90^{th} - 10^{th} Percentile (*r*)	0.60	0.12	0.85	0.78	0.35
Mean Bias (°)	-0.42	-0.52	-0.41	-0.42	-0.53
Lower Limit of Agreement (°)	-2.56	-3.19	-2.44	-2.27	-4.39
Upper Limit of Agreement (°)	1.72	2.16	1.61	1.44	3.33
Upper – Lower (°)	4.28	5.35	4.05	3.71	7.72
Axial Rotation					
Mean Angular Displacement (*r*)	---	---	---	---	0.15
90^{th} - 10^{th} Percentile (*r*)	---	---	---	---	0.73
Mean Bias (°)	---	---	---	---	0.18
Lower Limit of Agreement (°)	---	---	---	---	-4.95
Upper Limit of Agreement (°)	---	---	---	---	5.31
Upper – Lower (°)	---	---	---	---	10.26

* Column headers are defined in text and Table 1.

Discussion

Relatively small mean angular displacement RMSD estimates in the flexion/extension, lateral bending, and axial rotation motion planes were observed between the IMU system and the LMM. Strong correlation coefficients in the flexion/extension motion plane and small Bland and Altman bias estimates in the flexion/extension, lateral bending, and axial rotation motion planes were observed across a range of experimental conditions that included a variety of movements and work speeds. Although not directly comparable, RMSD estimates from this study were similar to those reported in studies comparing trunk motion measurements obtained with IMU-based instrumentation systems to OMC systems and anthropometry-corrected accelerometers (Goodvin et al., 2006; Kim and Nussbaum, 2013; O'Sullivan et al., 2012; Plamondon et al., 2007; Van Driel et al., 2013; Wong and Wong, 2008). Overall, the results suggest the IMU system examined in this study may serve as an acceptable instrument for directly measuring thoracolumbar trunk motion in field-based studies.

Errors in thoracolumbar trunk motion measures obtained during field-based assessments may vary based on the applications of interest (e.g., different work activities), characteristics of the individual direct measurement technology components (e.g., noise parameters of sensors), and methods used to estimate and/or summarize motion. One possible limitation of one sensor accelerometer or IMU methods is that trunk inclination with respect to gravity may not fully capture relevant trunk motion information. While some research has been conducted investigating the accuracy of two accelerometer systems (mounted over the sternum and sacrum) to assess trunk flexion (Koehoorn, 2010; Van Driel et al., 2009), it is unclear if methods using two accelerometer or IMU sensors may be a more appropriate for estimating "risk" of adverse health outcomes in comparison to trunk motion estimates from one sensor.

In this study, processing methods that computed thoracolumbar trunk motion as a function of measurements obtained from IMUs secured to both the sternum and L5/S1 body segments were more comparable to the LMM than processing methods that computed thoracolumbar trunk motion as a function of measurements obtained solely from the sternum mounted IMU. For example, the mean 90^{th} percentile angular displacement value estimated by the complementary weighting algorithm-based method (Comp-1) was nearly 15° greater than the estimate obtained with the LMM in the flexion/extension motion plane whereas the two sensor complementary weighting algorithm-based method (Comp-2) was within about 3° of the LMM. When considering that the 90^{th} percentile of angular displacement in the flexion/extension motion plane is commonly used as an estimate of the “peak” amount of trunk flexion in field studies aimed at estimating the association between exposure to non-neutral working postures and musculoskeletal outcomes, the results of this study suggest investigators should strongly consider computing thoracolumbar trunk motion as a function of estimates from multiple IMUs rather than using a single accelerometer secured to the sternum. However, future research examining the association between exposures to non-neutral working postures as measured with both one and two sensor methods and adverse health effects such a MSDs is necessary.

Another main finding of this study was that summary measures estimated with fusion algorithms such as the complementary weighting algorithm to combine gyroscope measurements with accelerometer measurements obtained from the IMUs agreed more strongly with summary measure estimates from the LMM than summary measures based solely on measurements from accelerometers. For example, the mean 90^{th} percentile estimates from the two IMU complementary weighting algorithm-based method (Comp-2) consistently agreed more strongly with the LMM than the two accelerometer method (Accel-2) for angular displacement, velocity, and acceleration in both the flexion/extension and lateral bending motion planes. The implication of this result is that use of IMU sensors and fusion algorithms may be an effective method for increasing the accuracy of accelerometer-based motion measurements that are known to be negatively affected by dynamic work processes (Amasay et al., 2009; Brodie et al., 2008a; Godwin et al., 2009; Hansson et al., 2001a).

Several limitations of this study should be acknowledged. First, although widely used in field studies, the LMM is not considered the "gold-standard" of trunk motion measurement. However, the objective of this study was not to compare IMU estimates of thoracolumbar trunk motion to a "gold-standard" OMC system. Rather, we compared two systems used in field-based studies where the IMU is less intrusive than the LMM. Mean angular displacement RMSD estimates may, therefore, be reduced or increased in comparison to an OMC system. Regardless, the conclusions regarding the use of two sensor IMU systems versus one sensor systems and the utility of the fusion algorithms hold. Strengths of this study include data collection across of a range of experimental conditions which allowed comparison of the IMU methods to the LMM across a variety of MMH task conditions. Additionally, the large number of participants (N=36) in comparison to previous, similar studies enhances generalizability and statistical stability.

The manufacturer provided Kalman-based estimate (HM Analyzer) was the only processing method that used the magnetometer measurements obtained with the IMUs in this study, and was therefore the only measurement method used to provide estimates of thoracolumbar trunk motion in the axial rotation motion plane. Performance of the Kalman-based method may have been affected by ferromagnetic disturbances in the laboratory environment or as a result of the proximity of the IMUs to the LMM during the experimental procedures. However, we visually inspected the calibration quality of the Kalman-based estimate after fitting both sensor systems and monitored signal quality during the MMH task using the HM Analyzer software and observed no evidence that such disturbances occurred. Thus, substantial performance degradation was not believed to have occurred.

While correlation coefficients assessing the linear relationship between the LMM and all of the IMU methods for mean angular displacement and angular displacement variation in the flexion/extension motion plane were strong, correlation coefficients in the lateral bending and axial rotation motion planes were generally weak to only moderately strong. The relatively poor performance of the IMU methods in the lateral bending and axial rotation motion planes are likely the result of a lack of variation between participants in the amount of lateral bending and axial rotation required by the MMH task. Much of the axial rotation completed by participants' to reach the box may be explained by reaching of the arms and rotation of the pelvis and trunk together.

The IMU system evaluated in this study produced estimates of trunk angular displacement that agreed reasonably well with analogous estimates from the LMM and thus is a promising alternative to the LMM for field-based studies. Several features of the IMU system, such as small size, wireless sensors, and data logging capability, are attractive from the perspective of obtaining high quality measurements of trunk motion in field-based research settings. Measurements obtained from IMUs secured to the sternum and pelvis had smaller root-mean-square differences and mean bias estimates in comparison to results obtained with the LMM than results of measurements obtained solely from a sternum mounted IMU. Additionally, fusion of IMU accelerometer measurements with IMU gyroscope measurements was observed to increase comparability to the LMM. Investigators should strongly consider computing thoracolumbar trunk motion as a function of estimates from multiple IMUs using fusion algorithms rather than using a single accelerometer secured to the sternum in field-based studies. Further exploration of fusion algorithms may improve the accuracy of IMU measurements for more complex joints such as the shoulder and/or wrist and documented field use of the IMU system under dynamic working conditions are needed.

CHAPTER III

ACCURACY AND REPEATABILITY OF AN INERTIAL MEASUREMENT UNIT SYSTEM FOR FIELD-BASED OCCUPATIONAL STUDIES

Introduction

Characterization of the association between non-neutral working postures and work-related musculoskeletal disorders (MSDs) requires accurate and precise posture measurement for optimal exposure assessment. Direct measurement methods are widely considered to provide the most precise and unbiased information content for estimating occupational exposure to physical risk factors for MSDs, in comparison to self-report or observation-based methods (Burdorf and van der Beek, 1999; David, 2005; Li and Buckle, 1999; Winkel and Mathiassen, 1994; Teschke et al., 2009; Trask et al., 2007; Vieira and Kumar, 2004). Accelerometers and gyroscopes, for example, are two small and portable direct measurement instruments commonly used in field-based studies to assess exposure to non-neutral working postures of the low back and shoulder (e.g., Fethke et al., 2011; Douphrate et al., 2012; Paquet, Punnett, and Buchholz, 2001; Teschke et al., 2009; Van Driel et al., 2013; Wong, Lee, and Yeung, 2009). Despite their common use, accelerometer-based estimates of posture have been observed to suffer from poor accuracy when work tasks involve complex, dynamic motions (Amasay et al., 2009, 2013; Brodie et al., 2008a; Godwin et al., 2009; Hansson et al., 2001a). Gyroscope-based estimates have been observed to suffer from large integration errors that severely restrict the duration of accurate measurements (Luinge and Veltink, 2005). These limitations have led investigators to consider alternative direct measurement technologies that may be better suited for field-based exposure assessment studies.

An inertial measurement unit (IMU) is a small and portable device that permits estimation of the spatial orientation of an object by combining the outputs of multiple electromechanical sensors (e.g., accelerometers and gyroscopes) through recursive sensor fusion algorithms such as a Kalman filter or complementary weighting algorithm (Bachmann et al., 1999; Gallagher et al., 2004; Higgins, 1975; Kalman 1960; Ligorio and Sabatini, 2013; Luinge et al., 1999; Luinge and Veltink, 2005; Sabatini 2006, 2011; Wagenaar et al., 2011; Yun and Bachmann, 2006). IMUs are considered advantageous to individual electromechanical sensors as fusion of orientation information from multiple electromechanical sensors may help overcome the limitations of each individual sensor component. For example, gyroscopic drift may be compensated for by fusing accelerometer-based orientation information resulting from the constant acceleration of gravity (Bachman et al., 2007; Favre et al., 2006; Luinge and Veltink, 2004; Wong and Wong, 2008; Zhou et al., 2006; Zhu and Zhou, 2004).

Several IMU systems have been observed to accurately estimate joint kinematics of the upper arm/shoulder (Cutti et al., 2008; de Vries et al., 2010; El-Gohary and McNames, 2012; Zhou et al., 2006, Zhou et al., 2007, 2008, 2010), the cervical spine (Duc et al., 2014; Theobald et al., 2012; Jasiewicz et al., 2007), the lower extremity (O'Donovan et al., 2009; Favre et al., 2008; Ferrari et al., 2009; Fong and Chan, 2010; Picerno et al., 2008), the trunk (Giansanti et al., 2007; Goodvin et al., 2006; Kim and Nussbaum, 2013; Lee et al., 2003; Roetenberg et al., 2007), and the whole body (Brodie et al., 2008b) in comparison to laboratory-based human motion analysis techniques such as optical motion capture (OMC) (Cuesta-Vargas et al., 2010). Despite their agreement with OMC systems in a laboratory setting, most studies examining the accuracy of IMU-based measurements have not sufficiently evaluated the repeatability of those measurements over a substantial time period, such as over the course of a full work shift (Mieritz et al., 2012). Some studies such as Plamondon et al. (2007), Kim and Nussbaum (2013), and Wong and Wong (2008), have had participants perform dynamic, intermediate duration tasks (lasting 30, 20, and 120 minutes in length, respectively) to combat this limitation in their performance evaluations of IMUs. Longer time frames are necessary, however, if IMUs are to be considered effective instruments for estimating occupational exposure to non-neutral postures associated with the development of MSDs in field-based studies.

The aim of this study was, therefore, to evaluate the accuracy and repeatability of estimates of trunk angular displacement in the flexion/extension and lateral bending motion planes and upper arm elevation (defined as either forward flexion or abduction of the upper arm) obtained with a commercially available IMU system over the course of an eight-hour work shift. The study was conducted in two phases: (1) a laboratory-based evaluation of the accuracy and repeatability of the IMU system in comparison to a gold-standard, OMC system, and (2) a field-based assessment of the repeatability of the IMU system during full work shift dairy parlor work, an occupation associated with substantial exposure to non-neutral postures and musculoskeletal health outcomes (Douphrate et al., 2009, 2012).

Methods

Laboratory Data Collection

To evaluate the accuracy and repeatability of the IMU system in a laboratory setting, a simulated milking cluster attachment task common to dairy parlor work was completed by one participant while trunk angular displacement angles in the flexion/extension and lateral bending motion planes and dominant upper arm elevation were simultaneously measured using two systems: (1) an eight-camera OMC system (Model: MX-40, Vicon Systems, Centennial, CO, USA), and (2) a commercially available IMU system (I2M Motion Tracking, Series SXT, Nexgen Ergonomics, Inc., Pointe Claire, Quebec, CAN).

The simulated cluster attachment task imitated a common, cyclic work task performed by dairy parlor workers in the real work environment. In the field, workers bend forward and toward the dominant arm to reach and grasp a milking cluster (hanging at waist height) with both hands and then lift and secure the cluster to the teats of a cow before repeating the task on the next cow in line.

To determine if the performance of the IMU system changed over time (e.g., due to “drift”), one “block” consisting of ten, simulated milking cluster attachment cycles was completed at the beginning of every hour for eight hours. The first block was considered a baseline measurement and was referred to as “Block 0”. Each block began with the participant standing in an upright stance, with the arms hanging relaxed, and the feet shoulder-width apart. At the start of each hour, the participant attached one milking cluster to a simulated cow teat. After the participant attached the milking cluster to the teat, he briefly returned to the resting position while a trained observer returned the milking cluster to its original starting location. Once the milking cluster was back in the starting position, the participant repeated the attachment task until the entire block of ten cycles had been completed. At the end of each block, the participant was allowed to rest in a chair while a trained investigator monitored marker and IMU placement to minimize the potential for marker or IMU movement errors.

The OMC reference system used single, passive reflective markers over the sternal notch, spinous process of the seventh cervical spine (C7), xyphoid process, acromion process, medial/lateral humeral epicondyle, anterior arm, radial/ulnar styloid process on the dominant limb, and on bilateral anterior superior iliac spine. Additionally, clusters of 3 markers were placed over the spinous process of the 8th thoracic spine (T8), sacrum, and over the 3rd metacarpal head on the dominant limb. The marker locations were selected based on the recommendation from the International Society of Biomechanics (ISB) (Wu et al., 2002, 2005). Marker data were initially digitized at 80 Hz and then down sampled to 20 Hz using linear interpolation to match trunk angular displacement and upper arm elevation information obtained with the IMU system.

The IMU system consisted of three units: one IMU was secured to the anterior torso at the sternal notch, one IMU sensor was secured to the posterior pelvis at the L5/S1 vertebrae, and one IMU was secured to the lateral aspect of the dominant upper arm approximately one-half the distance between the lateral epicondyle and the acromion. Specifically, the IMUs were placed into small, custom pockets that were sewn into a nylon and spandex triathlon suit that the participant wore while completing the task. Compression wrap was used to minimize potential IMU movement on the skin. The IMU data streams were sampled at 20 Hz and stored to on-board flash memory. The IMU data files were then downloaded to a desktop computer workstation and synchronized with the reference system recordings using a custom LabVIEW program. Additional details of the IMU specifications may be found in Chapter II.

Figure 3. Eight-camera OMC system and simulated milking cluster attachment task setup.

Figure 4. Participant performing milking cluster attachment task.

Field Data Collection

Field-based data were collected in milking parlors of three large-herd dairy operations during the summer months of 2014. These dairies were located in Colorado, New Mexico, and Texas. Among these three dairies were one parallel parlor, one herringbone parlor, and one rotary parlor. Ten dairy workers who each performed a full, eight hour work shift were recruited for this study. All participants were male (mean age=24 years, SD=1.8) and right-hand dominant. Participants had a median height of 1.6 m (range of 1.6-1.8 m), a median body mass of 69.9 kg (range of 63.5-81.6 kg), and a median body mass index of 27.2 kg/m2 (range of 25.6-30.0 kg/m2).

Approximately 45 min prior to starting work, each participant was fitted with three IMUs as described for the laboratory-based data collection procedure and a fourth IMU was placed on the non-dominant upper arm. Study procedures were approved by the University of Texas Health Science Center at Houston, Committee for the Protection of Human Subjects Institutional Review Board and written informed consent was obtained.

Instrumentation and Data Processing

Raw 3-dimensional coordinate data obtained with the OMC system (sampled at 80 Hz) were low-pass filtered (zero-phase, 4th order Butterworth; 17 Hz cutoff frequency) prior to down sampling to 20 Hz. The filtered and down sampled data were then used to calculate estimates of trunk angular displacement and upper arm elevation relative to the global coordinate system (OMC_Global). An OMC-based estimate of trunk angular displacement relative to the pelvis (OMC_Pelvis) and an estimate of upper arm elevation relative to the torso (OMC_Torso) were also calculated for comparison to IMU-based measures of trunk and shoulder motion, respectively. The anatomic coordinate systems of the pelvis, upper torso, and the shoulder joint were defined as recommended by the ISB (Wu et al., 2002, 2005). The shoulder joint center was defined as described by Rab et al. (2002) and shoulder angles were calculated using an Euler-Cardan angle method with rotation orders as recommended by the ISB (Wu et al., 2005). The upper torso orientation in the global reference frame was calculated using an Euler-Cardan angle method with a rotation order of flexion/extension, lateral bending, and axial rotation.

Four IMU processing methods were used to estimate trunk angular displacement in the flexion/extension and lateral bending motion planes for both the laboratory and field-based data collection portions of this study. The four methods included: 1) a low passed (zero-phase, 2nd order Butterworth; 3 Hz cutoff frequency) accelerometer-based estimate from the IMU secured to the sternum only (Accel-1); 2) a complementary weighting algorithm-based estimate incorporating accelerometer and gyroscope measurements from the IMU secured to the sternum only (Comp-1); 3) a low passed (zero-phase, 2nd order Butterworth; 3 Hz cutoff frequency) accelerometer-based estimate calculated as the difference of the estimates provided from the IMUs secured to the sternum and L5/S1 body segments (Accel-2); and 4) a complementary weighting algorithm-based estimate calculated as the difference of complementary-based estimates from the IMUs secured to the sternum and L5/S1 body segments (Comp-2). Estimates of trunk angular displacement in the axial rotation motion plane were not analyzed as ferromagnetic disturbances in both the laboratory and field environments (determined through visual inspection of the angular displacement waveforms during analysis) prevented use of the magnetometer readings in a Kalman-based estimate.

Three IMU processing methods were used to obtain estimates of dominant upper arm elevation for laboratory data collection and bilateral upper arm elevation for field-based data collection. The three methods included: 1) a low passed (zero-phase, 2nd order Butterworth; 3 Hz cutoff frequency) accelerometer-based estimate from the IMU secured to the arm only (Accel-1); 2) a complementary weighting algorithm-based estimate incorporating accelerometer and gyroscope measurements from the IMU secured to the arm only (Comp-1); and 3) a complementary weighting algorithm-based estimate calculated as the difference of complementary-based estimates from the IMUs secured to the sternum and the arm (ShoRT - "Shoulder Relative to Torso").

Accelerometer and custom complementary weighting algorithm-based estimates for the trunk and upper arm were derived as described in Chapter II, except that a weighting coefficient (K) value of 0.06 was used to compensate for the drift that occurs when a raw gyroscope signal is integrated (Luinge and Veltink, 2005) and accelerometer-based inclination angle estimates from the IMU secured to the arm were calculated as $\cos^{-1}(Ax / \sqrt{Ax^2 + Ay^2 + Az^2})$. The weighting coefficient of 0.06 was used to maintain the complementary weighting algorithm time constant of 0.77 sec evaluated in Chapter II, based on the weighting coefficient, the sampling rate (20 Hz), and the IMU gyroscope drift rate (approximately 1° s^{-1}). The inclination angle (α_n) was low pass filtered (zero-phase, 2^{nd} order Butterworth; 3 Hz cutoff frequency) and the angular velocity (ω_n) signals were high pass filtered (zero-phase, 2^{nd} order Butterworth; 0.5 Hz cutoff frequency) prior to computation of complementary inclination angle estimates.

Statistical Analysis

Estimates of the minimum, maximum, mean, 10^{th} percentile, 90^{th} percentile, 99^{th} percentile, and the difference between the estimates of the 90^{th} and 10^{th} percentiles (referred to as the angular displacement variation) were calculated from the angular displacement waveforms obtained from each IMU processing method and the OMC reference system for the laboratory-based analysis. Sample-to-sample root-mean-square differences (RMSD) for each block of cluster attachments were estimated by comparing the waveforms of each IMU processing method to the waveform obtained with the OMC reference system. The RMSD for each participant was calculated using Equation (2) (see Chapter II), where θ was the estimate from an IMU processing method, θ' was the estimate from the OMC reference system, n was the number of samples across the block of ten cluster attachment cycles, and i was sample number.

Linear regression was used to model the relationship between the changes in degrees of mean angular displacement and mean angular displacement variation (90th – 10th percentile) of trunk angular displacement in the flexion/extension and lateral bending motion planes and dominant upper arm elevation per eight hours increase in time. Slope estimates were used to assess the repeatability of the mean angular displacement and angular displacement variation for each IMU processing method for both the laboratory and field-based analysis.

Results

Laboratory-based Assessment of Accuracy

The OMC reference system and each of the IMU measurement methods produced waveforms of trunk angular displacement and upper arm elevation with relatively similar characteristics (Figures 5, 6, and 7). Estimates of the RMSD between the OMC system and the IMU measurement methods for the flexion/extension and lateral bending trunk motion planes and for upper arm elevation are provided in Tables 9, 10, and 11, respectively.

RMSD orientation error estimates between 5° and 7.5° were observed for all IMU processing methods in the flexion/extension and lateral bending trunk motion planes. RMSD estimates were generally similar (within 0.5 degrees) across methods obtained using only the IMU secured to the sternum and across methods obtained using both the IMUs secured to the sternum and L5/S1 body segment. However, the two IMU accelerometer-based (Accel-2) and complementary-based (Comp-2) methods used to estimate trunk angular displacement in the lateral bending motion plane as the difference of the estimates provided from the IMU secured to the sternum and L5/S1 body segments had a slightly larger RMSD (1.5 degrees). In general, the methods that used the IMUs secured only to the sternum were observed to be more comparable to the OMC system than methods obtained with IMUs secured to the sternum and L5/S1 body segment when considering RMSD orientation error estimates.

For the upper arm, RMSD orientation error estimates ranged from 7.3° for the accelerometer-based estimate from the IMU secured to the arm only (Accel-1) method to 12.4° for the complementary weighting algorithm-based estimate calculated as the difference of complementary-based estimates from the IMUs secured to the sternum and upper arm (ShoRT). The solely accelerometer-based estimate obtained from the IMU secured to the upper arm (Accel-1) had a smaller RMSD orientation error estimate in comparison to the OMC system than the complementary weighting algorithm-based estimate incorporating accelerometer and gyroscope measurements from the IMU secured to the upper arm only (Comp-1).

While RMSD estimates were generally smaller for solely accelerometer-based methods in comparison to complementary-based methods for both the trunk and shoulder, summary measure estimates such as the mean, 10th percentile, and 90th percentile angular displacement were more comparable with the OMC system for the complementary weighting algorithm-based methods that calculated the estimates of angular displacement using accelerometer and gyroscope measurements from the IMU secured to the sternum or upper arm only (Comp-1) than solely accelerometer-based estimates from the IMU secured to the sternum or upper arm (Accel-1).

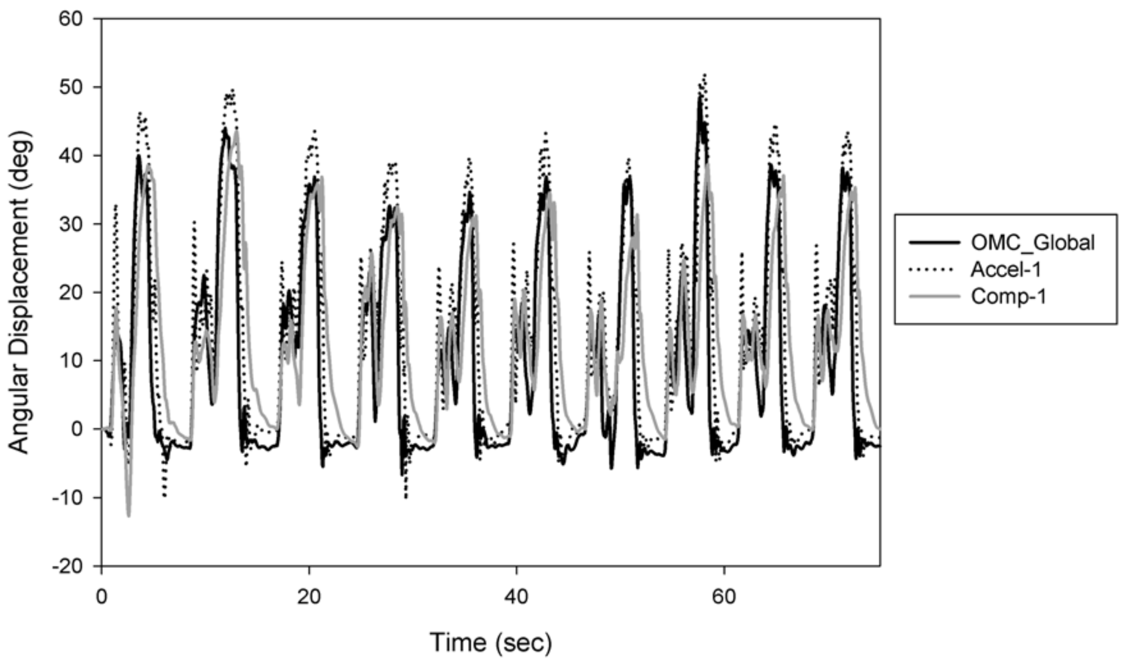

Figure 5. Upper arm elevation waveforms obtained with the OMC system and two IMU measurement processing methods for one block of the cluster attachment task.

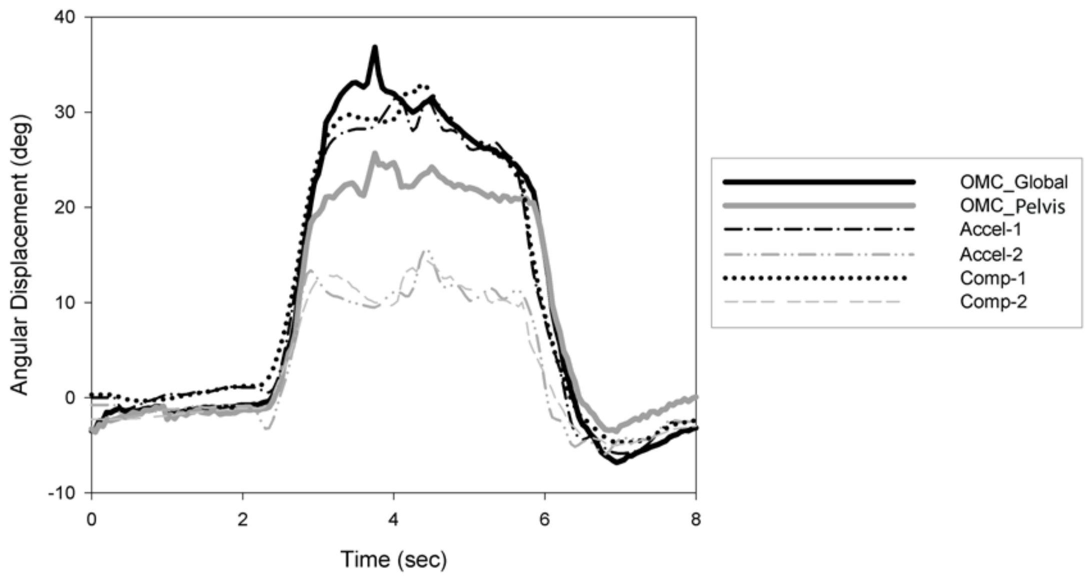

Figure 6. Trunk angular displacement waveforms obtained with the OMC system and the four IMU measurement methods in the flexion/extension motion plane for a single milking cluster attachment task cycle.

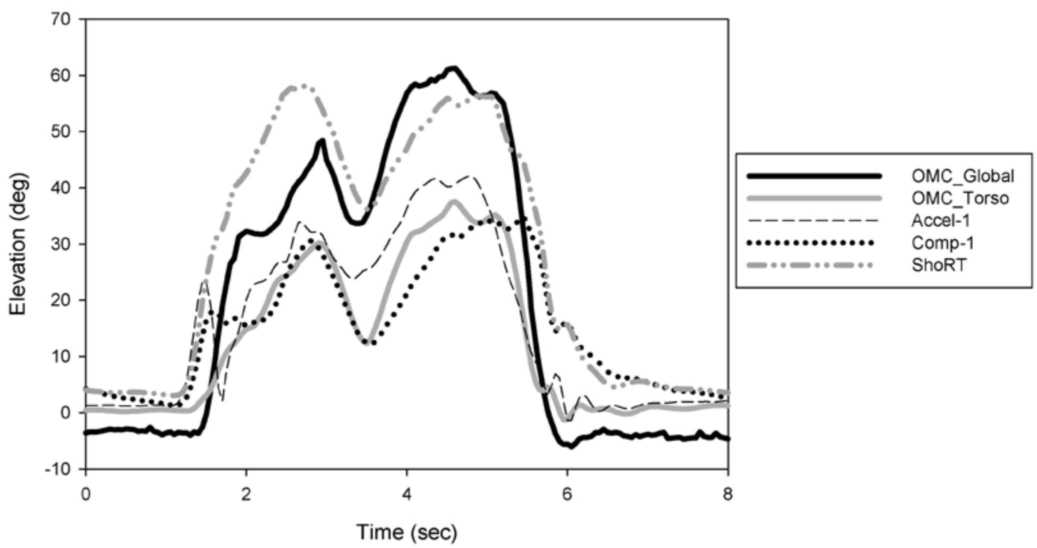

Figure 7. Upper arm elevation waveforms obtained with the OMC system and the three IMU measurement methods for a single milking cluster attachment task cycle.

Table 9. Mean (SD) of trunk angular displacement estimates in the flexion/extension motion plane by measurement method.*

Summary measure	OMC_Global	Accel-1	Comp-1	OMC_Pelvis	Accel-2	Comp-2
Maximum Extension (°)	-7.2 (3.7)	-5.6 (1.5)	-4.6 (1.9)	-7.3 (3.8)	-6.8 (1.4)	-4.0 (2.2)
Mean (°)	12.6 (2.5)	13.5 (2.7)	14.0 (3.0)	11.9 (1.8)	7.5 (2.7)	7.5 (2.8)
Maximum Flexion (°)	44.5 (7.9)	50.7 (8.3)	46.2 (9.0)	34.1 (3.9)	25.0 (5.5)	24.7 (4.4)
10^{th} Percentile (°)	-2.2 (2.2)	-0.7 (1.3)	-0.3 (1.5)	-2.2 (1.7)	-1.4 (1.8)	-1.3 (2.1)
90^{th} Percentile (°)	31.4 (4.7)	31.8 (5.6)	32.6 (5.9)	26.0 (2.6)	18.0 (3.6)	17.9 (3.6)
99^{th} Percentile (°)	41.3 (6.9)	44.8 (8.1)	43.4 (7.9)	31.9 (4.2)	22.5 (4.7)	22.4 (4.0)
Sample-to-Sample RMSD (°)	-- Ref --	5.1 (2.3)	5.0 (2.5)	-- Ref --	7.0 (2.6)	7.5 (1.5)

OMC_Global = Low passed (zero-phase, 4^{th} order Butterworth, 17 Hz cutoff) OMC-based estimate relative to the global coordinate system

OMC_Pelvis = Low passed (zero-phase, 4^{th} order Butterworth, 17 Hz cutoff) OMC-based estimate relative to the pelvis

* Additional column headers are defined in text and Table 1

Table 10. Mean (SD) of trunk angular displacement estimates in the lateral bending motion plane by measurement method.*

Summary measure	OMC_Global	Accel-1	Comp-1	OMC_Pelvis	Accel-2	Comp-2
Maximum to the Left (°)	-11.9 (6.0)	-19.3 (5.7)	-17.3 (8.8)	-6.0 (4.8)	-10.8 (1.8)	-9.2 (5.2)
Mean (°)	2.1 (2.2)	1.6 (1.8)	2.0 (2.1)	3.5 (2.1)	2.4 (1.0)	2.3 (1.8)
Maximum to the Right (°)	18.5 (7.0)	24.2 (6.6)	18.5 (5.7)	15.2 (6.4)	21.3 (7.2)	13.1 (4.2)
10^{th} Percentile (°)	-1.6 (2.3)	-3.1 (2.3)	-2.5 (2.7)	-0.5 (2.1)	-1.2 (1.0)	-1.0 (1.8)
90^{th} Percentile (°)	7.0 (3.7)	8.3 (3.0)	8.4 (3.0)	8.2 (3.8)	8.3 (2.5)	6.3 (2.3)
99^{th} Percentile (°)	14.1 (5.3)	17.9 (4.4)	16.1 (5.0)	12.7 (5.2)	14.8 (3.4)	10.3 (3.1)
Sample-to-Sample RMSD (°)	-- Ref --	4.2 (1.4)	4.5 (0.7)	-- Ref --	6.4 (1.4)	4.9 (1.6)

* Additional column headers are defined in text, Table 1, and Table 9

Table 11. Mean (SD) of the dominant (right) upper arm elevation estimates by measurement method.*

Summary measure	OMC_Global	Accel-1	Comp-1	OMC_Torso	ShoRT
Mean (°)	12.5 (2.1)	14.2 (2.1)	13.6 (2.6)	21.6 (1.3)	27.0 (4.3)
Maximum Elevation (°)	44.9 (3.2)	48.3 (2.9)	41.5 (3.6)	66.5 (5.1)	65.0 (7.5)
10th Percentile (°) [a]	-0.2 (1.7)	0.0 (1.5)	0.7 (2.8)	-2.0 (2.4)	2.8 (2.8)
90th Percentile (°)	35.6 (2.9)	38.2 (2.7)	31.8 (3.1)	54.2 (4.0)	52.0 (5.5)
99th Percentile (°)	42.5 (2.6)	46.0 (2.7)	38.4 (3.0)	62.5 (3.9)	61.1 (6.9)
Sample-to-Sample RMSD (°)	-- Ref --	7.3 (1.7)	8.3 (1.9)	-- Ref --	12.4 (2.2)

OMC_Torso = Low passed (zero-phase, 4th order Butterworth, 17 Hz cutoff) OMC-based estimate relative to the torso

ShoRT = a complementary weighting algorithm-based estimate calculated as the difference of complementary-based estimates from the IMUs secured to the sternum and upper arm (shoulder relative to the torso)

[a] Negative values denote extension behind the body

* Additional column headers are defined in text and Table 1

Laboratory-based Assessment of Repeatability

The IMU system produced relatively stable mean angular displacement and mean angular displacement variation (90^{th} – 10^{th} percentile) estimates of trunk motion in the flexion/extension (sagittal) and lateral bending (coronal) planes and dominant upper arm elevation (Figure 8). With the exception of the angular displacement variation slope estimate for the complementary weighting algorithm-based estimate calculated as the difference of complementary-based estimates from the IMUs secured to the sternum and the right arm (ShoRT), all trunk and upper arm elevation slope estimates were <5° of change in mean angular displacement and mean angular displacement variation per eight hours of data collection (Table 12). Furthermore, the majority of mean angular displacement and mean angular displacement variation slope estimates were observed to be <3° of change in angular displacement per eight hours of data collection.

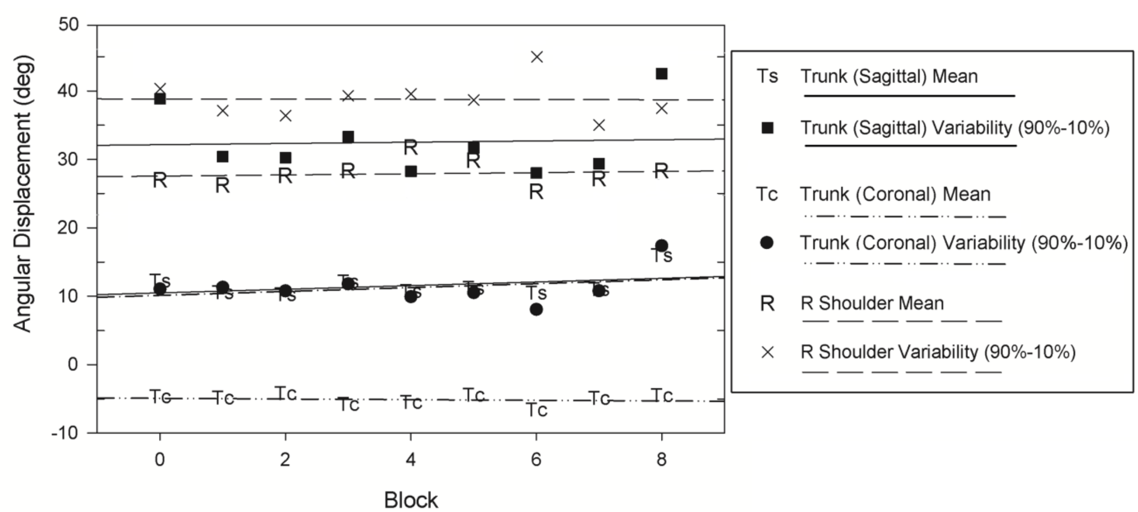

Figure 8. Trunk angular displacement and dominant (right) upper arm elevation estimates for the Accel-1 measurement method.

Table 12. Laboratory-based slope estimates (calculated as the ratio of the change in degrees of trunk angular displacement or upper arm elevation per eight hours) of the mean and angular displacement variation for each IMU processing method.

Summary Measure	Accel-1	Comp-1	Accel-2	Comp-2	ShoRT
Trunk (Flexion/Extension)					
Mean Angular Displacement	2.11	2.21	2.74	2.78	--
90^{th} - 10^{th} Percentile	0.72	0.14	1.01	1.30	--
Trunk (Lateral Bending)					
Mean Angular Displacement	-0.38	-0.34	-1.87	-1.82	--
90^{th} - 10^{th} Percentile	2.26	2.64	-2.26	-1.10	--
Right Shoulder					
Mean Angular Displacement	0.67	0.58	--	--	2.78
90^{th} - 10^{th} Percentile	-0.10	-1.78	--	--	-8.50
Left Shoulder					
Mean Angular Displacement	2.35	2.16	--	--	4.37
90^{th} - 10^{th} Percentile	-0.38	0.67	--	--	-3.31

* Column headers are defined in text, Table 1, and Table 11

Field-Based Assessment of Repeatability

Consistent with the results of the laboratory-based assessment of repeatability, the IMU system produced relatively stable mean angular displacement and mean angular displacement variation (90th – 10th percentile) estimates of trunk motion in the flexion/extension and lateral bending motion planes and bilateral upper arm elevation in the field (Figure 9). All trunk angular displacement and bilateral upper arm elevation slope estimates except for the angular displacement variation estimate for the complementary weighting algorithm-based estimate calculated as the difference of complementary-based estimates from the IMUs secured to the sternum and right arm (ShoRT) were observed to be <5° of change in mean angular displacement and mean angular displacement variation per eight hours of data collection (Table 13). Similar to the laboratory-based assessment, the majority of mean angular displacement and mean angular displacement variation slope estimates were observed to be <3° of change in angular displacement per eight hours of data collection.

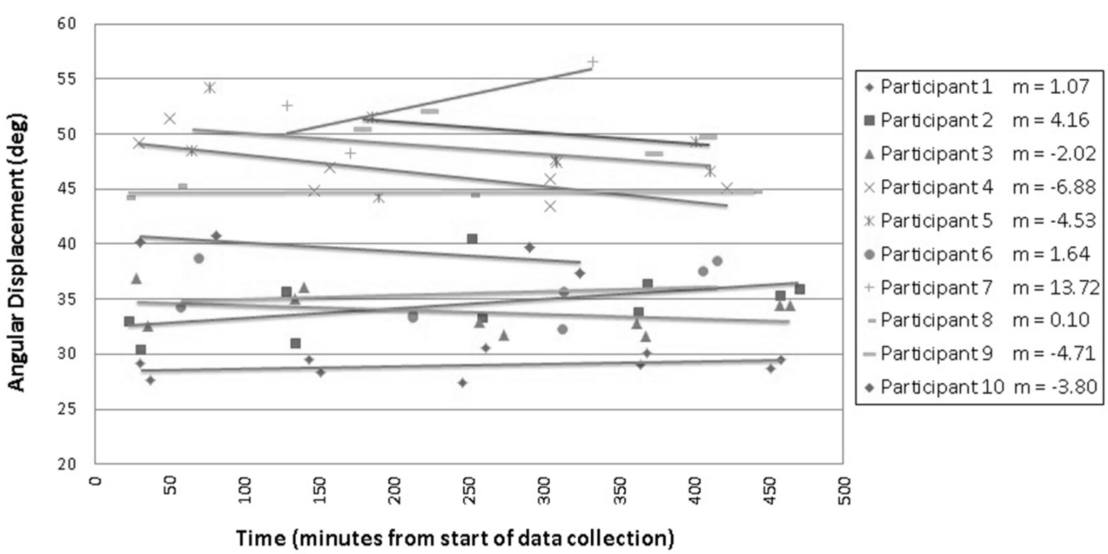

Figure 9. Left upper arm elevation and associated slope estimates for the Accel-1 measurement method for all participants.

Table 13. Field-based slope estimates (calculated as the ratio of the change in degrees of trunk angular displacement or upper arm elevation per eight hours) of the mean and angular displacement variation for each IMU processing method.

Summary Measure	Accel-1	Comp-1	Accel-2	Comp-2	ShoRT
Trunk (Flexion/Extension)					
Mean Angular Displacement	-2.88	-3.07	-1.92	-2.21	--
90^{th} - 10^{th} Percentile	-0.67	0.14	1.20	3.79	--
Trunk (Lateral Bending)					
Mean Angular Displacement	-0.24	-0.34	-1.10	-1.01	--
90^{th} - 10^{th} Percentile	0.24	1.97	1.01	1.01	--
Right Shoulder					
Mean Angular Displacement	3.74	4.37	--	--	0.86
90^{th} - 10^{th} Percentile	-1.82	-2.21	--	--	-6.86
Left Shoulder					
Mean Angular Displacement	-0.14	-0.10	--	--	-0.72
90^{th} - 10^{th} Percentile	-1.34	-3.12	--	--	0.72

* Column headers are defined in text, Table 1, and Table 11.

Discussion

The relatively small mean angular displacement RMSD estimates observed in the laboratory-based component of this study were similar to those reported in other laboratory-based studies comparing trunk motion and upper arm elevation measurements obtained with IMU-based instrumentation systems to OMC systems (Cutti et al., 2008; de Vries et al., 2010; El-Gohary and McNames, 2012; Goodvin et al., 2006; Kim and Nussbaum, 2013; Plamondon et al., 2007; Van Driel et al., 2013; Zhou et al., 2007, 2008, 2010). In particular, the RMSD estimates of 5° to 7.5° observed for all IMU processing methods in the flexion/extension and lateral bending trunk motion planes compare reasonably well with previous studies evaluating the accuracy of IMUs during dynamic, manual work tasks of intermediate durations (Kim and Nussbaum, 2013; Plamondon et al., 2007; Wong and Wong, 2008). This suggests the IMU system examined in this study may serve as an acceptable instrument for directly measuring flexion/extension and lateral bending trunk motion and upper arm elevation in field-based occupational exposure assessment studies.

While we generally observed smaller RMSD estimates for solely accelerometer-based methods in comparison to complementary-based methods for both the trunk and shoulder, peak summary measure estimates such as minimum, maximum, and 99th percentile angular displacement estimates for the complementary weighting algorithm-based methods that calculated the estimates of angular displacement using accelerometer and gyroscope measurements from the IMU secured to the sternum or upper arm only (Comp-1) were more comparable with the OMC system than the solely accelerometer-based method (Accel-1). For example, the maximum flexion estimate in the flexion/extension motion plane from the single sternum mounted IMU complementary weighting algorithm-based method (Comp-1) was observed to have an error of 1.7° in comparison to the OMC system than the one accelerometer method (Accel-1) that was observed to have an error of 6.2°. RMSD estimates may have been slightly larger for the complementary-based methods due to slight delay caused by the time constant (0.77 sec) of the recursive mathematical expression used to combine the accelerometer and gyroscope signals (Figure 5). The implication of this result is that while IMU sensors and fusion algorithms (i.e., Comp-1 and Comp-2) may not show substantially improved RMSD relative to gold-standard OMC systems in comparison to single-sensor approaches (i.e., Accel-1 and Accel-2), their use may improve the accuracy of estimates of peak or extreme postures during dynamic work processes with large ranges of postural excursion (Amasay et al. 2009; Brodie et al. 2008; Godwin et al. 2009; Hansson et al. 2001).

For both the laboratory and field-based assessments of repeatability, the majority of mean angular displacement and angular displacement variation trunk angular displacement and upper arm elevation slope estimates were observed to be <3° of change in mean angular displacement per eight hours of data collection. While we are aware of no empirical evidence suggesting acceptable amounts of IMU "drift" error, this level of error seems to be relatively minor and suggests that the performance of the IMU system evaluated in this study is rather stable and appropriate for long sampling durations (e.g., full work shift data collection).

This study represents the first effort we are aware of evaluating the accuracy and repeatability of estimates of trunk angular displacement and upper arm elevation obtained with a commercially available IMU system over the course of an eight-hour work shift in both a laboratory and field-based setting. The results contribute to the growing body of empirical evidence suggesting that IMUs may be used in field-based epidemiological studies to accurately assess exposure to occupational physical risk factors associated with musculoskeletal health outcomes.

The results of this study must be interpreted in the context of several limitations. Both the laboratory-based and field-based components of this study involved the analysis of a single cyclic work task performed by few participants (n=1 for laboratory component; n=10 for field component). The use of a single work task and few participants substantially limits the generalizability of the results to work tasks with comparable movement characteristics. Furthermore, while it was assumed that all participants performed the work task similarly in comparison to other participants and over time within a work shift, individual differences between study participants and uncontrolled factors such as fatigue and dairy parlor configuration likely affected the manner in which the work task was performed. These individual differences may partially explain changes in mean angular displacement and mean angular displacement variation (90^{th} – 10^{th} percentile) estimates over time.

Variable time between cluster attachment blocks and an inconsistent number of blocks performed by participants in the field-based component of this study may also have contributed to errors in the estimates of repeatability of trunk motion and upper arm elevation over time. For example, while most participants in the field-based component of this study performed between eight and ten blocks of the cluster attachment task during the course of their work day, participants working at the dairy parlor with a rotary style configuration were observed to perform only three to four cluster attachment blocks. Rotary style configurations generally employ a job rotation strategy that limits the number of cluster attachment blocks performed by a worker (Douphrate et al., 2012). The limited number of cluster attachment blocks may have created a situation in which some participants (for example, Participant 7 in Figure 9) had a larger slope than what might reasonably have been expected had more blocks had been performed.

Finally, it is unknown how additional work activities normally performed during dairy parlor work may have affected IMU accuracy in comparison to the OMC system. The participant in the laboratory-based component of this study was instructed to rest in a chair between blocks to preserve the location of OMC markers and IMU placement to ensure “drift” did not occur due to marker or IMU movement errors. While it is possible that IMU movement on the skin did occur, the use of the compression suit to minimize sensor movement appeared to work well and prevent gross changes in sensor placement. In a real work setting, additional work activities may contribute to increased errors in the accuracy of trunk motion and upper arm elevation estimates. Future work examining the performance of the IMU system with larger sample sizes and during completion of multiple work tasks with different kinematic characteristics (e.g. speeds and ranges of motion) is necessary. Additionally, further research on methods and strategies to improve the accuracy of the individual sensors that comprise an IMU would be of value.

CHAPTER IV

A COMPARISON OF EXAMINATION EQUIPMENT USED DURING COMMON CLINICAL OPHTHALMOLOGIC TASKS

Introduction

Work-related musculoskeletal disorders (MSDs) are prevalent among health care professionals (Waters et al., 2006; Ngan et al., 2010). Among health care and social assistance workers in the United States, MSDs accounted for 42% of non-fatal injuries and illnesses requiring days away from work with an incidence rate (55 cases per 10,000 full-time workers) higher than the rate for all private industries and second only to the transportation and warehousing industry (BLS, 2013). Ophthalmologists and other eye care physicians, in particular, report a high prevalence of musculoskeletal pain and other symptoms consistent with MSDs of the neck, shoulders, low back, and upper extremities (Chatterjee et al., 1994; Chams et al., 2004; Dhimitri et al., 2005; Marx et al., 2005; Long et al., 2011). Existing studies have reported prevalence estimates of neck symptoms ranging from 33% to 69% and upper extremity/shoulder symptoms ranging from 11% to 33% (Chams et al., 2004; Chatterjee et al., 1994; Dhimitri et al., 2005; Kitzmann et al., 2012; Sivak-Callcott et al., 2010). Recent evidence also suggests that eye care physicians report musculoskeletal pain of the neck and upper extremity more frequently than peers in other medical specialties (Kitzmann et al., 2012).

While the economic consequences of MSDs are substantial (Bhattacharya, 2014), data examining the costs of MSDs specific to ophthalmologists are currently unavailable. However, in a sample of 47 Australian optometrists, about 30% reported taking time off work while remaining in the profession and 45% reported seeking treatment for musculoskeletal pain “at least once every three months” (Long et al., 2014). Health care professionals in similar fields such as dentistry have also reported taking more sick leave, reducing their work hours, and even switching professions as a result of their musculoskeletal conditions (Akesson et al., 1999; Garbin et al., 2011; Osborn et al., 1990).

Several studies have reported associations between physical risk factors and MSDs among workers in occupations similar to ophthalmology, including dental hygienists (Hayes et al., 2009, 2013), hospital physicians (Hengel et al., 2011), and surgeons (Gofrit et al., 2008; Nimbarte et al., 2013; Sivak-Callcott et al., 2011; Stomberg et al., 2010; Szeto et al., 2009). Ophthalmologists may be at risk for developing MSDs due to their exposure to physical risk factors such as sustained muscular exertions and/or non-neutral working postures (van der Windt et al., 2000; Svendsen et al., 2004a, 2004b; Vieria and Kumar, 2004; da Costa and Vieira, 2010; Silverstein et al., 2008). As in laparoscopic surgery, ophthalmologists are often challenged by having restricted access to the patient, a limited ability to reposition their equipment, and the need to simultaneously focus instruments while manipulating controls (Berguer et al., 1999; Matern, 2009; van Veelen et al., 2004). However, unlike laparoscopic surgery (Berguer and Smith, 2006; Matern et al., 2009; Trejo et al., 2007; Van der Schatte Olivier et al., 2009), little empirical information is available to assist practitioners in the identification and control of exposures to physical risk factors in the ophthalmic clinical environment.

Two clinical instruments commonly used during eye examinations that may expose ophthalmologists to physical risk factors are the slit-lamp biomicroscope and the binocular indirect ophthalmoscope. A conventional slit-lamp biomicroscope is operated by looking through viewing oculars with 0° of inclination (with respect to horizontal) while adjusting the instrument’s position and focus using one or both hands. This task often requires an ophthalmologist to sustain a position of non-neutral neck flexion to obtain a clear view into the patient’s eye through the viewing oculars. Ophthalmologists will also frequently hold an external lens up to the patient’s eye, requiring prolonged periods of upper arm elevation.

The binocular indirect ophthalmoscope is commonly used while the patient lies supine or sits upright in the examination chair, depending on the personal preferences of the ophthalmologist and/or positioning restrictions of the patient. The ophthalmologist moves around the patient to obtain views into the eye from various angles using a handheld lens. Use of a conventional binocular indirect ophthalmoscope may result in exposure to prolonged periods of neck flexion, trunk flexion, and upper arm elevation. Furthermore, the design of a conventional binocular indirect ophthalmoscope may require elevated levels of neck and shoulder muscle activity to support the weight of the device, which is often concentrated on the ophthalmologist's forehead.

Alternative ophthalmologic examination equipment has recently become available that has the potential to reduce exposure to physical risk factors in the clinical environment. This alternative equipment includes a slit-lamp biomicroscope with wider table-top, inclined ocular adaptors, adjustable height elbow supports, a pneumatic examination stool with adjustable body support, and a wireless binocular indirect ophthalmoscope. The wider slit-lamp biomicroscope table-top and elbow supports provide an ophthalmologist a means to more comfortably rest the arms during use of the instrument and may minimize shoulder discomfort or fatigue, while the inclined oculars may promote less biomechanically stressful neck postures. The stool can be adjusted to provide support of the arms or back depending on the preference of the ophthalmologist and may provide an additional method for resting the arms during a clinical examination. In comparison to the conventional binocular indirect ophthalmoscope, the wireless binocular indirect ophthalmoscope has a more even weight distribution, which may reduce biomechanical loading of the neck/shoulder region. The wireless capability of the binocular indirect ophthalmoscope also allows for more access to the patient.

The objective of this study was to compare the levels of muscle activity, trunk inclination, and upper arm elevation associated with the use of conventional and alternative examination equipment during common clinical ophthalmologic tasks. In particular, we hypothesized that the alternative examination equipment would reduce the overall demands of the upper trapezius and anterior deltoid muscles during a common ophthalmologic exam in comparison to the conventional equipment. We also hypothesized that the alternative examination equipment would lead to a reduction in the percentage of time spent working with the upper arms elevated in comparison to the conventional equipment.

Methods

Participants

Two mock clinical examinations were conducted by a convenience sample of 15 ophthalmologists (10 male, 5 female; all right hand dominant), which included a combination of 9 faculty, 2 fellows, and 4 residents (mean age=41.9 years, SD=11.9), and who were recruited from the University of Iowa Hospitals and Clinics (UIHC) Department of Ophthalmology. Participants reported no history of physician-diagnosed MSDs in the neck/shoulder region and no episodes of neck/shoulder pain within 14 days prior to participation. All study procedures were approved by the University of Iowa Institutional Review Board and informed consent was obtained prior to participation.

Participants had a median height of 1.8 m (range of 1.6-1.9 m), a median body mass of 68.1 kg (range of 56.8-104.4 kg), and a median body mass index of 23.3 kg/m2 (range of 21.5-32.1 kg/m2). The participants reported a median of 10 years of clinical experience (range of 2-36 years). Fifteen potential participants were excluded based on either self-reported histories of physician-diagnosed MSDs in the neck/shoulder region or self-reported episodes of neck/shoulder pain within 14 days prior to expressing interest in participating.

Mock Clinical Examinations

Participants performed one mock clinical examination in a room with conventional equipment available in all examination rooms at the UIHC ophthalmology clinic (i.e., the conventional condition), and one mock clinical examination in a second room with alternative equipment (i.e., the alternative condition). Patients for the mock clinical examinations used an identical script and were coached on how to perform. Each patient presented a chief complaint of floaters in both eyes for the past one year, with no change in the quantity, no flashes of lights, and no vision changes. Floaters are undissolved gel particles that occasionally float in the liquid center of the vitreous humour, the thick fluid or gel that fills the eye. Floaters are typically the result of natural aging. All patients had both eyes dilated. Participants were instructed to examine the patient as they normally would in the clinic environment for both room conditions (i.e., they were not instructed to examine the patient in any particular manner or order), except that they must complete all clinical tasks in each room. Participants were also instructed to use the binocular indirect ophthalmoscope in each room with the patient in the same position (i.e., patient lied supine or sat upright in each room condition).

Clinical tasks performed by each participant in each examination room included: 1) initial patient interview and associated “documentation” (i.e., using computer or completing paperwork), 2) fitting the patient to the slit-lamp biomicroscope, 3) examining the patient using the slit-lamp biomicroscope without the use of a handheld lens, 4) examining the patient using the slit-lamp biomicroscope with the use of a handheld lens in the right hand, 5) examining the patient using the slit-lamp biomicroscope with the use of a handheld lens in the left hand, 6) putting away the slit-lamp biomicroscope, 7) putting on the indirect ophthalmoscope, 8) examining the patient with the indirect ophthalmoscope, 9) removing the indirect ophthalmoscope, and 10) patient exit interview and associated documentation.

In the room with the conventional equipment, use of personal equipment such as a handheld lens case or the slit lamp biomicroscope as a means to support the arms during the exams was permitted. In the room with the alternative equipment, participants were instructed to use the provided alternative equipment. A block randomization procedure was used to counterbalance the order of room presentation and digital video recordings were obtained for each mock examination. The conventional examination room was equipped with a pneumatic examination stool (Reliance model 1020B), a slit-lamp biomicroscope with straight (0° of inclination) viewing oculars (Haag Streit PN 900.7.2.6989), and a binocular indirect ophthalmoscope (Heine Omega 180).

The alternative examination room was equipped with a pneumatic examination stool with adjustable body support (Reliance model 5346), a slit-lamp table top (Reliance Xoma) with adjustable elbow supports for use while examining a patient with a handheld lens, a slit-lamp biomicroscope with inclined ocular adaptors (Haag Streit PN 09007.8), and a wireless binocular indirect ophthalmoscope (Keeler Vantage Plus Wireless PN 1205P1020) (Figure 10). Participants were initially unfamiliar with the equipment in the alternative examination room. Thus, each participant was given time to inspect and test the alternative equipment prior to beginning study procedures. Features of the alternative equipment that differed from the conventional equipment were highlighted, and any questions participants had were addressed. Following each mock examination, participants completed a short usability survey regarding the equipment in that particular examination room.

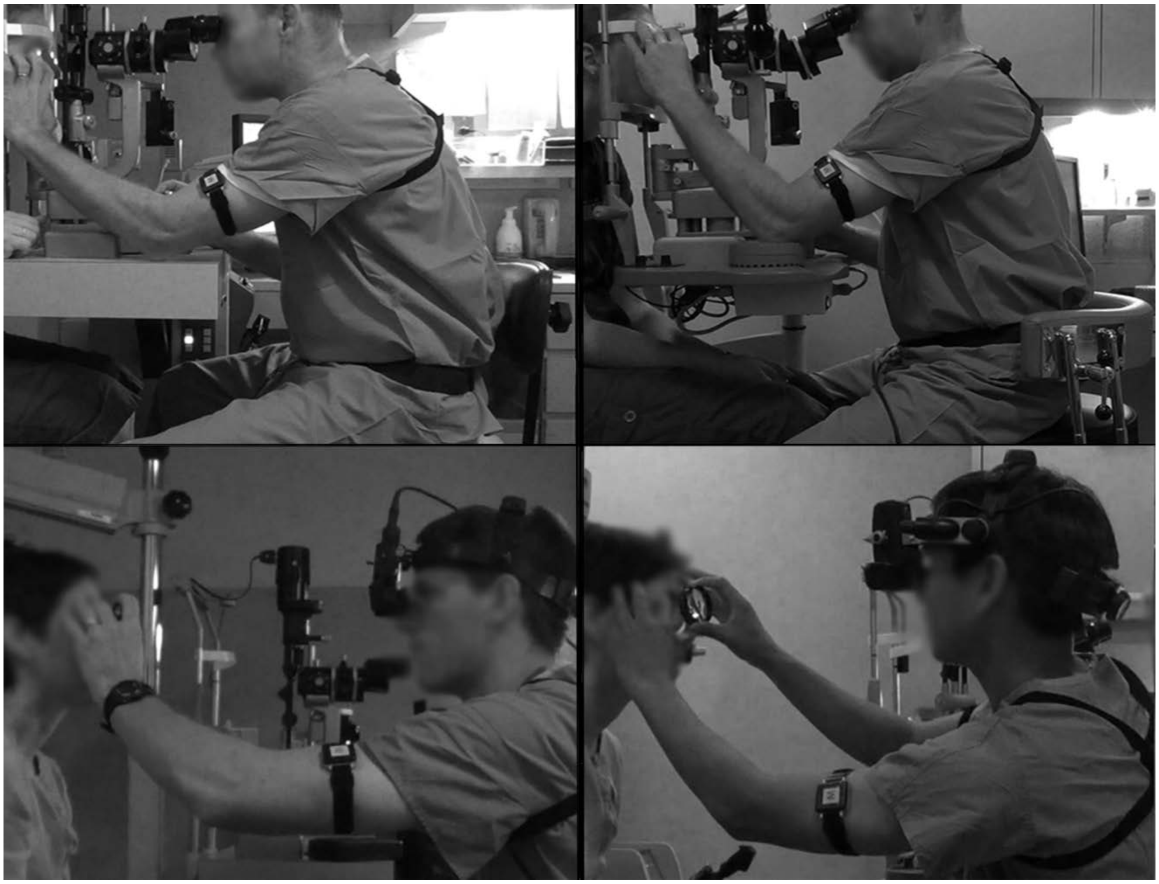

Figure 10. Participants examining a mock patient. Top Left: Conventional examination condition slit-lamp biomicroscope table with straight (0° of inclination) oculars, no elbow supports, and a non-adjustable stool. Top Right: Alternative examination condition slit-lamp biomicroscope table with inclined oculars, padded elbow supports, and pneumatic stool with adjustable body support. Bottom Left: Conventional examination condition binocular indirect ophthalmoscope. Bottom Right: Alternative examination condition wireless binocular indirect ophthalmoscope.

Instrumentation and Data Processing

Surface electromyography (EMG) was used to record bilateral myoelectric activity of the upper trapezius and anterior deltoid muscles. For the upper trapezius muscles, preamplified surface EMG electrodes (model DE2.3, Delsys Inc., Boston, MA) were secured to the skin slightly lateral of the midpoint between the acromion and the 7th cervical vertebra. Electrodes for the deltoid muscles were located approximately 4 cm below the midpoint between the acromion and the deltoid tubercle of the clavicle (Criswell, 2010). The electrodes had dual, bipolar, 10×1mm silver bars, an inter-electrode distance of 10 mm, differential amplification with a gain of 1000, and a 20-450 Hz bandwidth. A reference electrode was placed over the non-dominant clavicle. The electrodes were connected to a surface EMG instrumentation amplifier (Bagnoli-16, Delsys Inc., Boston, MA), and the raw EMG signals were sampled at 1000 Hz and stored to a desktop computer workstation for signal processing and analysis. Final electrode placement was verified by examining EMG signal quality during manually-resisted isometric contractions.

All EMG recordings were processed and analyzed with custom LabVIEW (version 2013, National Instruments, Inc., Austin, TX) and Matlab (r2013b, The MathWorks, Inc., Natick, MA) software. Electrocardiogram artifacts, transient artifacts, and other potential sources of interference (e.g., 60 Hz) were managed using procedures described previously (Fethke et al., 2011). Each raw EMG recording was then converted to instantaneous root-mean-square (RMS) amplitude using a 100-sample moving window with a 50-sample overlap.

The RMS EMG amplitudes recorded during the mock examinations were normalized as a percentage of the RMS EMG amplitude observed during submaximal, isometric reference contractions (%RVE). Submaximal contractions were used instead of maximal voluntary contractions (MVC) because participants may have difficulty generating maximum contractions in a time efficient manner (Mathiassen et al. 1995;

Hägg et al., 1997), and to minimize the risk of discomfort and injury during normalization procedures (Nieminen et al. 1993; Mathiassen et al. 1995; Attebrant et al. 1995; Bao et al., 1995).

For the upper trapezius, reference contractions were obtained while participants held a 2 kg weight in each hand with the upper arms elevated to 90° in the frontal plane (i.e., humeral abduction), elbows fully extended and forearms pronated (Mathiassen et al., 1995). For the anterior deltoid, participants held a 2 kg weight in each hand with the upper arms flexed forward to 90° of elevation and the elbows fully extended (Cook et al., 2004; Yoo et al., 2010; Rota et al., 2013). Three repetitions of each submaximal reference contraction were performed, with a 1-min rest period between repetitions. Participants maintained each submaximal reference contraction for 15 sec and the mean RMS amplitude of the middle 10 sec was calculated. For each muscle separately, the average of the mean RMS EMG amplitudes of the three reference contractions was used as the RVE activation level. A baseline RMS EMG amplitude level was also measured by having participants sit in a relaxed posture with the upper back and arms supported for 60 s. The baseline level was defined as the lowest RMS amplitude during the 60 s recording period and was quadratically subtracted from all subsequent RMS EMG amplitude values (Thorn et al., 2007).

Angular displacements of the thoracolumbar region of the trunk in the flexion/extension and lateral bending motion planes and upper arm elevation angles with respect to the gravity vector were estimated using four inertial measurement units (IMU) (I2M Motion Tracking, Series SXT, Nexgen Ergonomics, Inc., Pointe Claire, Quebec, CAN). IMU specifications and placement locations for the trunk were consistent with those described in Chapter II. Additionally, one IMU was secured to the skin of the lateral aspect of each upper arm approximately one-half the distance between the lateral epicondyle and the acromion.

The custom complementary weighting algorithm described in Chapter II was used to transform the raw IMU data into trunk angular displacement and upper arm elevation angles. In this study, upper arm elevation refers to either forward flexion or abduction of the upper arm. The complementary weighting algorithm approach was used in lieu of a solely accelerometer-based approach as the accuracy of accelerometer-based estimates have been observed to be less accurate under dynamic working conditions (Amasay et al. 2009; Brodie et al., 2008a; Godwin et al., 2009; Hansson et al. 2001a). Details of the mathematics of complementary weighting are found elsewhere (Higgins, 1975; Wagenaar et al., 2011; El-Gohary and McNames, 2012).

Summary Measures

The duration of each mock examination and of each clinical task within each mock examination was calculated through use of an event marker (digitized simultaneously with the surface EMG recordings) and the digital video recordings. Summary measures of normalized RMS EMG, trunk inclination, and upper arm elevation recordings were calculated across each entire mock examination and separately for each clinical task within each mock examination. For surface EMG, the arithmetic mean of the normalized RMS EMG amplitude (in %RVE) was calculated for each muscle. For trunk inclination and upper arm elevation, posture categories were used to describe percent time with trunk inclination >60º, with trunk lateral inclination (right or left) >15º, and with upper arm elevation >60º (Douphrate et al., 2012; Hooftman et al., 2009; Wahlström et al., 2010). The usability survey assessed participants’ perceptions of the slit lamp biomicroscope, indirect binocular ophthalmoscope, and the pneumatic stool regarding attributes such as comfort and adjustability. Participants’ ratings of equipment attributes were obtained using discrete 0-to-5 point scales with verbal anchors at 0 (poor) and 5 (excellent).

Statistical Analysis

Each mock examination was parsed according to clinical task as described above, with an entire exam comprising all tasks. Postural data were successfully obtained for all participants. For one participant, surface EMG data were lost due to instrumentation failure. Means and standard deviations were calculated for each summary measure and for each clinical task by examination room condition. Paired t-tests (two-tailed) were used to compare the muscle activity, trunk inclination, and upper arm elevation summary measures between the examination rooms. Comparisons of each clinical task between the examination room conditions were planned a priori, therefore, no adjustment was made for multiple comparisons (i.e., each comparison was evaluated for statistical significance using a p-value of 0.05). The Wilcoxon signed rank test was used to compare results of the equipment usability surveys between the examination rooms. All statistical analyses were conducted with SAS, version 9.3 (SAS Institute, Cary, NC).

Results

Mock Examination Duration

In general, we observed longer full examination and clinical task durations during the alternative equipment condition in comparison to the conventional equipment condition (Table 14). Statistically significant differences were observed for the clinical tasks of fitting a patient to the slit lamp biomicroscope, slit lamp biomicroscope use when not holding a lens, and indirect ophthalmoscope use. Although the average duration of the full mock examination was longer for the alternative equipment condition in comparison to the conventional equipment condition, the difference was not statistically significant.

Table 14. Mean (SD) of clinical task duration (sec) by examination room condition.

Clinical Task	Conventional	Alternative	p [a]
Entire Exam	461.0 (135.4)	514.9 (114.9)	0.07
Initial Interview Documentation	88.9 (40.1)	88.5 (40.6)	0.97
Fitting Patient to Slit Lamp	27.0 (15.6)	38.7 (16.9)	**0.01**
Slit Lamp Use (No Lens)	29.4 (18.9)	37.4 (18.6)	**0.03**
Slit Lamp Use (Lens R Hand)	28.9 (19.7)	33.1 (15.4)	0.19
Slit Lamp Use (Lens L Hand)	32.2 (25.4)	36.1 (19.3)	0.45
Putting Away Slit Lamp	21.8 (24.3)	20.1 (13.0)	0.79
Putting On Indirect Lamp	37.0 (16.8)	34.7 (13.8)	0.38
Indirect Lamp Use	89.0 (74.9)	99.7 (79.8)	**0.03**
Putting Away Indirect Lamp	20.4 (16.2)	19.3 (6.2)	0.75
Exit Interview Documentation	86.3 (48.8)	107.2 (43.6)	0.09

[a] *p*-values obtained from paired t-tests

Surface Electromyography

Across an entire exam, the alternative equipment condition resulted in small (<4% RVE) but statistically significant reductions in the average mean RMS upper trapezius EMG amplitude in comparison to the conventional equipment condition (Table 15). The alternative equipment condition also resulted in a statistically significant reduction in the average mean RMS amplitude of the left anterior deltoid across an entire exam in comparison the conventional equipment condition (Table 16).

The alternative equipment condition resulted in reductions in muscle activity levels in comparison to the conventional equipment condition for the majority of clinical tasks and muscle groups examined. In most cases, the reductions were small and not statistically significant. However, use of the alternative slit lamp biomicroscope while holding an external lens to the patient's eye was associated with substantial reductions in right and left upper trapezius muscle activity and in right anterior deltoid activity. The observed reductions depended on the hand in which the external lens was held (i.e., reduced right upper trapezius mean RMS amplitude when holding the external lens with the right hand).

For the clinical task of initial interview documentation, the alternative equipment condition resulted in a statistically significant reduction in the average mean RMS amplitude of the right upper trapezius in comparison to the conventional equipment condition. Additionally, the alternative equipment condition resulted in a statistically significant reduction in the average mean RMS amplitude of the left anterior deltoid in comparison to the conventional equipment condition for the clinical task of exit interview documentation.

Table 15. Mean (SD) of mean normalized root-mean-square surface electromyography amplitudes [a] by clinical task and examination room condition [b] for the Upper Trapezius muscles.

Clinical Task	Right Upper Trapezius			Left Upper Trapezius		
	Conventional	Alternative	*p*	Conventional	Alternative	*p*
Entire Exam	26.3 (4.9)	22.6 (6.9)	**0.01**	27.8 (9.1)	25.1 (10.8)	**<0.005**
Initial Interview Documentation	18.9 (10.0)	14.5 (7.1)	**<0.005**	15.2 (10.0)	14.3 (10.8)	0.51
Fitting Patient to Slit Lamp	35.3 (14.4)	41.3 (13.1)	0.21	38.2 (16.8)	41.7 (13.6)	0.14
Slit Lamp Use (No Lens)	39.1 (23.7)	34.8 (15.8)	0.32	42.0 (30.1)	40.2 (22.9)	0.57
Slit Lamp Use (Lens R Hand)	24.0 (17.3)	10.9 (7.5)	**0.02**	35.1 (22.9)	17.4 (10.2)	**<0.005**
Slit Lamp Use (Lens L Hand)	29.8 (20.9)	18.4 (14.3)	**0.01**	22.4 (14.1)	13.7 (11.9)	**0.03**
Putting Away Slit Lamp	27.8 (10.5)	29.4 (14.1)	0.69	33.6 (10.6)	28.9 (14.6)	0.22
Putting On Indirect Lamp	39.5 (15.3)	37.7 (15.3)	0.62	37.0 (15.9)	33.2 (12.1)	0.36
Indirect Lamp Use	28.1 (15.7)	29.6 (23.4)	0.68	39.8 (28.0)	40.5 (30.2)	0.81
Putting Away Indirect Lamp	35.4 (11.2)	32.5 (14.2)	0.57	35.4 (12.9)	33.3 (14.1)	0.42
Exit Interview Documentation	19.0 (11.6)	14.2 (5.4)	0.20	17.5 (12.4)	12.7 (6.6)	0.14

[a] Expressed as percentage of reference voluntary exertion (%RVE)

[b] *p*-values obtained from paired t-tests for N = 14 participants

Table 16. Mean (SD) of mean normalized root-mean-square surface electromyography amplitudes [a] by clinical task and examination room condition [b] for the Anterior Deltoid muscles.

Clinical Task	Right Anterior Deltoid			Left Anterior Deltoid		
	Conventional	Alternative	*p*	Conventional	Alternative	*p*
Entire Exam	19.9 (9.6)	19.0 (9.6)	0.41	22.6 (8.8)	20.9 (8.1)	**0.03**
Initial Interview Documentation	9.6 (6.5)	10.1 (8.3)	0.65	20.4 (14.2)	17.7 (9.9)	0.27
Fitting Patient to Slit Lamp	20.8 (7.9)	22.6 (8.5)	0.20	23.4 (7.5)	27.1 (9.9)	0.10
Slit Lamp Use (No Lens)	17.6 (14.0)	16.1 (7.8)	0.56	23.1 (11.6)	25.2 (10.2)	0.31
Slit Lamp Use (Lens R Hand)	27.2 (20.5)	15.5 (11.5)	**0.05**	17.3 (14.1)	13.0 (7.9)	0.12
Slit Lamp Use (Lens L Hand)	20.2 (14.5)	13.5 (8.4)	**0.01**	22.8 (16.4)	20.5 (20.1)	0.70
Putting Away Slit Lamp	22.2 (15.7)	25.6 (16.1)	0.16	23.6 (15.2)	23.5 (10.3)	0.98
Putting On Indirect Lamp	28.6 (10.2)	28.7 (14.5)	0.95	29.7 (14.8)	26.8 (10.4)	0.34
Indirect Lamp Use	40.1 (34.5)	40.8 (42.3)	0.81	33.1 (14.9)	32.5 (15.8)	0.77
Putting Away Indirect Lamp	32.1 (28.4)	28.5 (13.5)	0.61	27.4 (22.4)	23.7 (14.6)	0.53
Exit Interview Documentation	12.0 (6.8)	10.8 (5.6)	0.46	16.7 (9.4)	13.0 (6.3)	**0.01**

[a] Expressed as percentage of reference voluntary exertion (%RVE)

[b] *p*-values obtained from paired t-tests for N=14 participants

Upper Arm Elevation and Trunk Inclination

For the upper arms, the alternative equipment condition generally resulted in small, non-statistically significant reductions in the percentage of time elevated >60° in comparison to the conventional equipment condition (Table 17). However, a statistically significant reduction in the percentage of time with the left upper arm elevated >60° was observed for the clinical task of holding an external lens to the patient's eye with the left hand. Conversely, we observed a statistically significant increase in the percentage of time with the left upper arm elevated >60° when fitting the patient to the slit lamp biomicroscope in the alternative equipment condition in comparison to the conventional equipment condition. No statistically significant differences in the percentage of time with trunk inclination >60° or lateral inclination >15° were observed for the alternative equipment condition in comparison to the conventional equipment condition.

Table 17. Mean (SD) of mean percent of time with shoulder elevated by clinical task and examination room condition [a].

Clinical Task	Right Upper Arm Elevation			Left Upper Arm Elevation		
	Conventional	Alternative	*p*	Conventional	Alternative	*p*
Entire Exam	15.6 (9.7)	12.9 (7.1)	0.13	16.0 (7.1)	15.3 (8.9)	0.66
Initial Interview Documentation	1.9 (3.5)	1.9 (2.1)	0.93	1.6 (3.3)	1.2 (3.5)	0.42
Fitting Patient to Slit Lamp	9.7 (12.0)	6.3 (6.7)	0.18	15.6 (15.3)	24.9 (13.0)	**<0.005**
Slit Lamp Use (No Lens)	6.0 (12.4)	2.4 (3.2)	0.23	12.0 (14.0)	17.9 (21.2)	0.21
Slit Lamp Use (Lens R Hand)	71.3 (32.2)	52.4 (44.6)	0.10	10.8 (21.4)	5.4 (19.3)	0.49
Slit Lamp Use (Lens L Hand)	9.1 (7.2)	7.7 (18.2)	0.73	76.5 (18.7)	52.7 (39.9)	**0.03**
Putting Away Slit Lamp	14.3 (25.2)	10.4 (15.6)	0.50	2.6 (4.3)	3.5 (5.1)	0.52
Putting On Indirect Lamp	24.8 (12.8)	22.4 (12.2)	0.44	18.8 (13.6)	19.8 (18.8)	0.84
Indirect Lamp Use	32.8 (35.1)	31.5 (37.4)	0.54	38.2 (32.9)	35.7 (36.1)	0.63
Putting Away Indirect Lamp	15.5 (20.9)	17.1 (14.0)	0.74	18.1 (25.5)	10.2 (7.5)	0.23
Exit Interview Documentation	3.3 (7.1)	1.5 (2.3)	0.41	0.8 (1.7)	1.0 (2.4)	0.79

[a] *p*-values obtained from paired t-tests for N=15 participants

Usability

For all equipment attributes examined, participants rated the alternative examination equipment more favorably than the conventional examination room equipment (Table 18). Statistically significant differences were observed for ease of moving and adjustability of the slit lamp biomicroscope, comfort and adjustability of the indirect lamp, and comfort and adjustability of the pneumatic stool.

Table 18. Mean (SD) of examination room equipment attribute ratings

Attribute	Conventional	Alternative	*p* [a]
Slit Lamp			
Ease of moving	3.3 (0.8)	4.3 (0.8)	**<0.01**
Comfort	3.4 (0.8)	3.8 (0.8)	0.25
Adjustability	3.0 (0.8)	4.1 (0.8)	**<0.01**
Indirect Lamp			
Comfort	2.9 (1.0)	4.7 (0.5)	**<0.01**
Adjustability	2.9 (1.2)	4.6 (0.5)	**<0.01**
Pneumatic Stool			
Control	3.8 (0.8)	3.8 (1.1)	0.84
Comfort	3.5 (0.5)	4.0 (0.9)	**0.05**
Adjustability	3.3 (0.7)	3.9 (1.0)	**0.05**

[a] *p*-values obtained from Wilcoxon signed rank tests for N=15 participants

Discussion

Occupational exposure to sustained muscular exertions and non-neutral working postures has been associated with the development of musculoskeletal pain and other symptoms consistent with MSDs in healthcare workers (Waters et al., 2006; Ngan et al., 2010). In particular, working with the upper arms elevated in positions >60° may be hazardous as the space between the humeral head and the acromion narrows such that pressure on the supraspinatus tendon is greatest (NIOSH, 1997). The increased pressure may lead to degenerative changes of the tendons of the rotator cuff, predisposing workers to tears (Armstrong et al., 1993; Svendsen et al., 2004a, 2004b). Improving the ergonomics of the equipment used in the clinical environment has been suggested as one method of minimizing exposures to risk factors such as sustained muscular exertion and non-neutral postures experienced by ophthalmologists and other eye care physicians (Marx, 2012). The alternative examination equipment used in this study may reduce the muscular effort required of the upper trapezius and right anterior deltoid muscles during a common ophthalmologic exam. Furthermore, the results suggest that use of the alternative examination equipment may reduce the percentage of work time spent with the upper arms elevated in positions >60º during use of the slit lamp biomicroscope while holding an external lens.

Specifically, the clinical task of using the alternative slit lamp biomicroscope while holding an external lens resulted in substantial reductions in upper trapezius (both right and left) and right anterior deltoid EMG amplitude and the percentage of time with the left upper arm elevated >60° in comparison to use of the conventional slit lamp biomicroscope. Participant use of elbow supports on the extended slit-lamp table top while examining a patient with a handheld lens may explain the reductions. Previous research during sedentary and manual work has suggested that arm supports may lead to improvements in subjective comfort and reductions of muscle activity in the shoulder and upper extremity (Milerad et al., 1994; Feng et al., 1997; Odell et al., 2007). Haddad et al.

(2012) observed reductions of upper trapezius activity to <5% of a maximum voluntary contraction when an ergonomically designed chair with arm supports was used by dentists.

Review of the video recordings showed that 8 of the 15 participants used either their personal handheld lens case (n=2) or the slit lamp biomicroscope table itself (n=6) to support their arms while examining the patient using the slit-lamp biomicroscope with the use of a handheld lens in the conventional equipment condition. The elbow supports provided in the alternative equipment condition were used by all participants. Exploratory comparisons suggested that muscle activity levels among participants using a lens case or the slit lamp biomicroscope table-top for elbow support during the conventional equipment condition were not meaningfully different than muscle activity levels observed during the alternative equipment condition (data not shown). Thus, the reductions in muscle activity levels observed during use of alternative slit lamp biomicroscope among the full study sample appear to be strongly influenced by absence of elbow supports among some participants during the conventional equipment condition.

Use of the alternative binocular indirect ophthalmoscope examined in this study did not result in any statistically significant reductions of muscle activity or percentage of time with the upper arms elevated. Despite these findings, statistically significant differences in participants’ ratings of the comfort and adjustability of the binocular indirect ophthalmoscope were observed suggesting that the alternative equipment was preferred by the ophthalmologists in comparison to the conventional equipment. Additional examination of the potential biomechanical advantages of use of the alternative binocular indirect ophthalmoscope is recommended.

There were several limitations to this study. First, the study sample was a convenience sample of predominantly experienced ophthalmologists in an academic setting who may or may not have adjusted their behaviors over time in response to the conditions of their work environment. The non-random selection of participants may have led to sample distortion, making the findings less generalizable to the entire population of ophthalmologists including those in private practice.

For the clinical task of initial interview documentation, the alternative equipment condition resulted in a statistically significant reduction in the average mean RMS amplitude of the right upper trapezius in comparison to the conventional equipment condition. Additionally, for the clinical task of exit interview documentation, the alternative equipment condition resulted in a statistically significant reduction in the average mean RMS amplitude of the left anterior deltoid in comparison to the conventional equipment condition. Reductions for these clinical tasks were not expected as all documentation equipment such as the computer used by the ophthalmologists was identical in both conditions. Differences in mean RMS EMG amplitude between the conventional and alternative equipment conditions may have occurred as a result of differences in communication styles between participants. In particular, some participants engaged in lengthy conversation (e.g., >2 minutes in duration) with the patients during the clinical task of exit interview documentation, which occurred more frequently during the alternative equipment condition than during the conventional equipment condition. Review of video recordings obtained during experimental procedures suggested that when such conversations occurred, participants were sitting with the arms generally relaxed. Therefore, longer exit interview documentation durations during the alternative equipment condition likely led to longer periods of low EMG activity and the lower mean RMS EMG amplitudes observed for each muscle (although the difference was statistically significant only for the left anterior deltoid).

For the clinical task of fitting a patient to the slit lamp biomicroscope, a statistically significant increase in the duration of the task was observed between the conventional and alternative room conditions. We anticipate that this difference was a result of the ophthalmologists' unfamiliarity with the alternative equipment and recognize that this limitation may have affected our estimates of average mean RMS muscle activity and percentage of time with the upper arm elevated >60°. In particular, a statistically significant increase in the percentage of time with the left upper arm elevated >60° was observed for this clinical task, revealing a potential trade-off in the use of the alternative equipment. However, estimates of the average mean RMS amplitude for fitting the patient to the slit lamp biomicroscope in the alternative equipment condition were not statistically significant for the four muscles examined in comparison to the conventional equipment condition. We suspect that the percentage of time with the left upper arm elevated >60° is likely an artifact of the increased duration fitting the patient to the slit lamp biomicroscope, which, with increased familiarity to the equipment, would decrease.

While the mock clinical examinations used in this study provided a stable test environment for comparisons of the conventional and alternative examination conditions, they may have removed many of the complexities ordinarily observed in a live clinical environment such as occupational psychosocial stress. Previous studies of musculoskeletal outcomes among ophthalmologists have observed positive associations between stress levels and the prevalence of neck, upper extremity, and lower back symptoms (Dhimitri et al., 2005; Kitzmann et al., 2012). Since information about occupational psychosocial stress was not collected in this study, it is unknown if the generally positive effects of the alternative examination equipment examined will transfer to the live environment.

Although the sample size of 15 ophthalmologists was sufficient for detecting differences in several EMG and posture summary measures, the clinical relevance of the observed reductions are unknown. For example, despite observing statistically significant reductions in upper trapezius (both right and left) and right anterior deltoid EMG amplitude during use of the alternative slit lamp biomicroscope while holding an external lens, the reductions of muscular exertion for this clinical task may not lead to the reduction of musculoskeletal symptoms among Ophthalmologists. Future work should examine similar clinical equipment prospectively, where associations between musculoskeletal outcomes and exposure to physical risk factors such as muscular exertion and non-neutral postures may be estimated following extended use of both conventional and alternate equipment configurations. Finally, non-neutral postures of the neck (e.g., protraction/retraction and flexion/extension) may also contribute to biomechanical loading during clinical ophthalmologic practice, particularly during use of the slit lamp biomicroscope. Future studies evaluating examination equipment commonly used in clinical ophthalmologic practice should attempt to characterize exposure to non-neutral postures of the neck.

To the best of our knowledge, this is the first study comparing estimates of muscle activity, trunk inclination, and upper arm elevation during use of conventional and alternative examination equipment for common clinical ophthalmologic tasks. While recommendations about ideal clinician positions during use of the slit-lamp biomicroscope and binocular indirect ophthalmoscope are available (Woolley and Kitzmann, 2011), no published, empirical evidence exists to support them or compare them with alternative instruments. This study contributes results from which practitioners can construct initial, evidence-based recommendations for the prevention of musculoskeletal symptoms and disorders among the broader population of ophthalmologic specialists.

CHAPTER V

CONCLUSIONS

Epidemiological evidence suggests an association between exposure to non-neutral working postures and work-related musculoskeletal disorders (MSDs) of the low back and shoulder. Direct measurement devices commonly used to assess occupational exposure to non-neutral working postures in field-based studies such as electrogoniometers and individual body-mounted electromechanical sensors (e.g., accelerometers and gyroscopes) may be obtrusive, often lack the portability necessary for field use, and/or have limited accuracy when used to measure dynamic or complex motions, especially over extended durations. Inertial measurement units (IMUs) are emerging instrumentation devices that measure and report an object's orientation and motion characteristics using multiple body-mounted electromechanical sensors. They have been observed to accurately monitor body kinematics of the trunk, shoulder, and other body regions over periods of relatively short duration in comparison to laboratory-based optical motion capture (OMC) systems. Until now, however, limited research has been performed to evaluate the accuracy of exposure information obtained with IMUs in comparison to exposure information obtained with other field-capable direct measurement devices. Additionally, insufficient information on the repeatability of IMU-based estimates over a substantial time period (e.g., a full work shift), and inadequate knowledge regarding the effects of different IMU sensor configurations and processing methods on the accuracy and repeatability of estimates of exposure has existed.

The work presented in this thesis was conducted to address these limitations and to provide novel information regarding the use of IMUs as direct measurement devices for assessing occupational exposure to non-neutral working postures. Three specific aims were developed and addressed through the completion of three studies:

SA #1: Evaluate the accuracy and repeatability of a commercially available IMU system for quantifying exposures to non-neutral trunk and shoulder postures for use in field-based occupational studies.

SA #2: Explore the effect of several different sensor configurations and processing methods on the estimates of exposure to non-neutral trunk and shoulder postures obtained with the IMU system.

SA #3: Apply the IMU system in a field-based occupational study to estimate exposures to non-neutral trunk and shoulder postures to demonstrate the utility of IMUs as direct measurement instruments.

The first study, presented in Chapter II, was performed to compare estimates of thoracolumbar trunk motion obtained with a commercially available IMU system with estimates of thoracolumbar trunk motion obtained with a field-capable reference system, the LMM, during completion of a simulated manual material handling task. A secondary objective of the study was to explore the effect of alternative sensor configurations and processing methods on the agreement between LMM and IMU-based estimates of trunk motion. The objectives of this study addressed specific aims one and two of the thesis. Results indicated that the IMU system exhibited reasonably good accuracy in comparison to the LMM, suggesting it may serve as an acceptable instrument for directly measuring thoracolumbar trunk motion in field-based studies. Processing methods that computed thoracolumbar trunk motion as a function of measurements obtained from IMUs secured to both the sternum and L5/S1 body segments were more comparable to the LMM than processing methods that computed thoracolumbar trunk motion as a function of measurements obtained solely from the sternum mounted IMU. Additionally, use of fusion algorithms that combined information obtained from the gyroscope and/or magnetometer components of the IMU sensors were observed to be an effective method for increasing the accuracy of accelerometer-based motion measurements that are known to be negatively affected by dynamic work processes. The primary implication of this study was investigators should consider computing thoracolumbar trunk motion as a function of estimates from IMUs secured to both the sternum and L5/S1 body segments using sensor fusion algorithms rather than using a single accelerometer secured to the sternum in field-based occupational exposure assessment studies.

Chapter III presented a study performed to evaluate the accuracy and repeatability of estimates of trunk angular displacement and upper arm elevation obtained with the IMU system examined in Chapter II over a long sampling duration. The effects of alternative sensor configurations and processing methods on the accuracy and repeatability of estimates of trunk angular displacement and upper arm elevation obtained with the IMU system were also explored. The study objectives addressed specific aims one and two of the thesis and involved two phases: (1) a laboratory-based evaluation of the accuracy and repeatability of the IMU system in comparison to a gold-standard, OMC system, and (2) a field-based assessment of the repeatability of the IMU system during full work shift dairy parlor work, an occupation associated with substantial exposure to non-neutral postures and musculoskeletal health outcomes. Results of the study suggested the IMU system evaluated in this thesis was reasonably accurate in comparison to an OMC system and displayed good repeatability for use in field-based occupational exposure assessment studies. In particular, results suggested the IMU system can be expected to exhibit <3° of IMU "drift" error over the course of an eight-hour work shift.

A randomized, repeated measures intervention study was presented in Chapter IV that demonstrated the use of the IMU system evaluated in Chapters II and III as a useful direct measurement device for comparing conventional and alternative, "ergonomic" examination equipment commonly used in clinical ophthalmologic practice. The study objective addressed specific aim three of the thesis. Results of the study suggested that some aspects of the alternative equipment evaluated in the study may help reduce exposures to sustained, non-neutral working postures of the neck and shoulder among ophthalmologists. More central to the context of this thesis, the study confirmed the utility of IMUs as a means to estimate exposure to non-neutral working postures associated with MSDs in a field setting.

Overall, the results presented in this thesis provide an original contribution to the scientific literature concerning best practices for assessing exposure to non-neutral working postures in field environments. Investigators may use the knowledge gained through these studies to more confidently use IMUs to estimate associations between exposure to non-neutral working postures and musculoskeletal health outcomes in a variety of high-risk occupational settings. It is suggested that future work focus on the application of IMUs to other body regions commonly affected by musculoskeletal pain (e.g., neck and distal upper extremities) and in studies examining the use of IMUs as a means to simultaneously measure multiple components of total worker health and well-being. For example, IMUs are theoretically capable of simultaneously measuring physical activity while obtaining more accurate information on exposure to non-neutral working postures than standard physical activity monitors. Simultaneous measurement of physical activity and working posture information would allow for a more complete characterization of the physical demands of many work activities and occupations than is currently available.

Further research on the fusion algorithms used to combine measurements from the individual sensors that comprise an IMU is needed. Some recent studies, for example, have placed particular focus on improving the methods used to implement magnetometer measurements into IMU-based measures and should continue as technological advancement permits increased sensor capabilities (Brodie et al., 2008b; Ettelt et al., 2014; Jiang et al., 2012; Schepers et al., 2010). Furthermore, research exploring the optimal weighting coefficients of sensor fusion algorithms (e.g., the complementary weighting algorithm) and studies investigating the use of IMUs to complement established observational exposure assessment methods, such as the Strain Index or

Revised NIOSH Lifting Equation would allow for more accurate and precise objective assessments of worker risk.

Finally, recent studies have suggested increasing motor variability as a potential strategy for supporting reductions of exposure to physical risk factors associated with MSDs at work (Mathiassen, 2006; Srinivasan and Mathiassen, 2012; Srinivasan et al., 2014). Extrinsic, postural feedback from wearable technologies such as IMUs may provide a means to train workers to increase motor variability (Ribeiro et al., 2011; 2013; 2014). Longitudinal field studies are needed, however, to confirm the existence of a potential relationship between motor variability and MSDs and to determine if more variation in biomechanical exposure is an effective intervention strategy (Srinivasan et al., 2014). Additional research on the different mechanisms by which postural feedback may be presented to increase motor variability (e.g., frequency, form, temporality, etc.) and the effects of the feedback over time are also needed (Ribeiro et al., 2011; 2014).

REFERENCES

Aggarwal, J. K., and Cai, Q. (1999). Human Motion Analysis: A Review. Computer Vision and Image Understanding, 73(3), 428-440.

Airaksinen, O., Brox, J., Cedraschi, C. o., Hildebrandt, J., Klaber-Moffett, J., Kovacs, F., Mannion, A.F., Reis, S., Staal, J.B., Ursin, H., Zanoli, G. (2006). Chapter 4 European guidelines for the management of chronic nonspecific low back pain. European Spine Journal, 15, s192-s300.

Åkesson, I., Johnsson, B., Rylander, L., Moritz, U., and Skerfving, S. (1999). Musculoskeletal disorders among female dental personnel—clinical examination and a 5-year follow-up study of symptoms. International Archives of Occupational and Environmental Health, 72(6), 395–403.

Altun, K., Barshan, B., and Tuncel, O. (2010). Comparative study on classifying human activities with miniature inertial and magnetic sensors. Pattern Recognition, 43, 3605–3620.

Amasay, T., and Karduna, A. (2013). Patient's Body Size Influences Dental Hygienist Shoulder Kinematics. IIE Transactions on Occupational Ergonomics and Human Factors, 1(3), 153-165.

Amasay, T., Zodrow, K., Kincl, L., Hess, J., and Karduna, A. (2009). Validation of tri-axial accelerometer for the calculation of elevation angles. International Journal of Industrial Ergonomics, 39(5), 783-789.

American Federation of Labor and Congress of Industrial Organizations (AFL-CIO) (2012). Report on 'Death on the Job, the Toll of Neglect: a National and State-by-state Profile of Worker Safety and Health in the United States'.

Aminian, K., Najafi, B., Büla, C., Leyvraz, P. F., and Robert, P. (2002). Spatio-temporal parameters of gait measured by an ambulatory system using miniature gyroscopes. Journal of biomechanics, 35(5), 689-699.

Andriacchi, T. P., and Alexander, E. J. (2000). Studies of human locomotion: past, present and future. Journal of Biomechanics, 33(10), 1217-1224.

Armstrong, T. J., Buckle, P., Fine, L. J., Hagberg, M., Jonsson, B., Kilbom, A., Silverstein, B.A., Sjogaard, G., and Viikari-Juntura, E. R. (1993). A conceptual model for work-related neck and upper-limb musculoskeletal disorders. Scandinavian journal of work, environment and health, 73-84.

Attebrant, M., Mathiassen, S. E., and Winkel, J. (1995). Normalizing upper trapezius EMG amplitude: Comparison of ramp and constant force procedures. Journal of Electromyography and Kinesiology, 5(4), 245–250.

Bao, S., Howard, N., Spielholz, P., Silverstein, B., and Polissar, N. (2009). Interrater reliability of posture observations. Human Factors: The Journal of the Human Factors and Ergonomics Society, 51(3), 292-309.

Bao, S., Mathiassen, S. E., and Winkel, J. (1995). Normalizing upper trapezius EMG amplitude: Comparison of different procedures. Journal of Electromyography and Kinesiology, 5(4), 251–257.

Bachmann, E.R. Inertial and magnetic tracking of limb segment orientation for inserting humans into synthetic environments, Ph.D. dissertation, Naval Postgraduate School, Monterey, CA, 2000.

Bachmann, E. R. (1997). Inertial and magnetic tracking of limb segment orientation for inserting humans into. Teleoperators and Virtual Environments, 6(1), 29-56.

Bachmann, E. R., Duman, I., Usta, U. Y., Mcghee, R. B., Yun, X. P., and Zyda, M. J. (1999). Orientation tracking for humans and robots using inertial sensors. In Computational Intelligence in Robotics and Automation, 1999. CIRA'99. Proceedings. 1999 IEEE International Symposium on (pp. 187-194). IEEE.

Baldwin, M. L. (2004). Reducing the costs of work-related musculoskeletal disorders: targeting strategies to chronic disability cases. Journal of Electromyography and Kinesiology, 14(1), 33-41.

Barriera-Viruet, H., Sobeih, T. M., Daraiseh, N., and Salem, S. (2006). Questionnaires vs observational and direct measurements: a systematic review. Theoretical Issues in Ergonomics Science, 7(3), 261-284.

Berguer, R., Forkey, D., and Smith, W. (1999). Ergonomic problems associated with laparoscopic surgery. Surgical Endoscopy, 13(5), 466–468.

Berguer, R., and Smith, W. (2006). An ergonomic comparison of robotic and laparoscopic technique: The influence of surgeon experience and task complexity. Journal of Surgical Research, 134(1), 87–92.

Bernmark, D., and Wiktorin, C. (2002). A triaxial accelerometer for measuring arm movements. Applied Ergonomics, 33: 541-547.

Bhattacharya, A. (2014). Costs of occupational musculoskeletal disorders (MSDs) in the United States. International Journal of Industrial Ergonomics, 44(3), 448-454.

Bland, J. M., and Altman, D. G. (1995). Comparing methods of measurement: why plotting difference against standard method is misleading. The lancet, 346(8982), 1085-1087.

Bland, J. M., and Altman, D. G. (1999). Measuring agreement in method comparison studies. Statistical methods in medical research, 8(2), 135-160.

Bland, J. M., and Altman, D. G. (2010). Statistical methods for assessing agreement between two methods of clinical measurement. International Journal of Nursing Studies, 47(8), 931-936.

Bland, M., and Altman, D. (1986). Statistical methods for assessing agreement between two methods of clinical measurement. The lancet, 327(8476), 307-310.

Boden, L. I., and Ozonoff, A. (2008). Capture–Recapture Estimates of Nonfatal Workplace Injuries and Illnesses. Annals of Epidemiology, 18(6), 500–506.

Brodie, M. A., Walmsley, A., and Page, W. (2008a). Dynamic accuracy of inertial measurement units during simple pendulum motion. Comput Methods Biomech Biomed Engin, 11(3), 235-242. doi: 10.1080/10255840802125526

Brodie, M. A., Walmsley, A., and Page, W. (2008b). Fusion motion capture: a prototype system using inertial measurement units and GPS for the biomechanical analysis of ski racing. Sports Technology, 1(1), 17–28.

Buchholz, B., and Wellman, H. (1997). Practical operation of a biaxial goniometer at the wrist joint. Human Factors, 39(1), 119-129.

Burdorf, A., and van der Beek, A. (1999). Exposure assessment strategies for work-related risk factors for musculoskeletal disorders. Scandinavian Journal of Work, Environment, Health, 25 Suppl 4, 25–30.

Bureau of Labor Statistics. (2011). Nonfatal occupational injuries and illnesses requiring days away from work, 2010. News (USDL-11-1612). Washington, DC: U.S. Department of Labor.

Bureau of Labor Statistics. (2012). Nonfatal occupational injuries and illnesses requiring days away from work, 2011. News (USDL-12-2204). Washington, DC: U.S. Department of Labor.

Bureau of Labor Statistics. (2013). Nonfatal occupational injuries and illnesses requiring days away from work, 2012. News (USDL-13-2257). Washington, DC: U.S. Department of Labor.

Catalfamo, P., Ghoussayni, S., and Ewins, D. (2010). Gait event detection on level ground and incline walking using a rate gyroscope. Sensors, 10(6), 5683-5702.

Cavuoto, L. A., and Nussbaum, M. A. (2014). Influences of Obesity on Job Demands and Worker Capacity. Current Obesity Reports, 1-7.

Chams, H., Mohammadi, S. F., and Moayyeri, A. (2004). Frequency and assortment of self-reported occupational complaints among Iranian ophthalmologists: A preliminary survey. Medscape General Medicine, 6(4).

Chatterjee, A., Ryan, W., and Rosen, E. (1994). Back pain in ophthalmologists. Eye, 8(4), 473–474.

Cook, C., Burgess-Limerick, R., and Papalia, S. (2004). The effect of upper extremity support on upper extremity posture and muscle activity during keyboard use. Applied Ergonomics, 35(3), 285–292.

Criswell, E. (2010). Cram’s introduction to surface electromyography: Chapter 4. Electrodes and site selection strategies. Sudbury, MA: Jones and Bartlett Publishers.

Coenen, P., Kingma, I., Boot, C. R., Twisk, J. W., Bongers, P. M., and van Dieën, J. H. (2013). Cumulative low back load at work as a risk factor of low back pain: A prospective cohort study. Journal of Occupational Rehabilitation, 23(1), 11-18.

Cuesta-Vargas, A. I., Galán-Mercant, A., and Williams, J. M. (2010). The use of inertial sensors system for human motion analysis. Physical Therapy Reviews, 15(6), 462-473.

Cutti, A. G., Giovanardi, A., Rocchi, L., Davalli, A., and Sacchetti, R. (2008). Ambulatory measurement of shoulder and elbow kinematics through inertial and magnetic sensors. Medical and biological engineering and computing, 46(2), 169-178.

da Costa, B. R., and Vieira, E. R. (2010). Risk factors for work-related musculoskeletal disorders: a systematic review of recent longitudinal studies. American journal of industrial medicine, 53(3), 285-323.

Dale, A. M., Strickland, J., Gardner, B., Symanzik, J., and Evanoff, B. A. (2010). Assessing agreement of self-reported and observed physical exposures of the upper extremity. International journal of occupational and environmental health, 16(1), 1-10.

Dartt, A., Rosecrance, J., Gerr, F., Chen, P., Anton, D., and Merlino, L. (2009). Reliability of assessing upper limb postures among workers performing manufacturing tasks. Applied ergonomics, 40(3), 371-378.

David, G. C. (2005). Ergonomic methods for assessing exposure to risk factors for work-related musculoskeletal disorders. Occupational Medicine, 55(3), 190-199.

de Vries, W. H. K., Veeger, H. E. J., Cutti, A. G., Baten, C. T. M., and van der Helm, F. C. T. (2010). Functionally interpretable local coordinate systems for the upper extremity using inertial and magnetic measurement systems. Journal of biomechanics, 43(10), 1983–1988.

Delisle, A., Larivière, C., Plamondon, A., and Imbeau, D. (2006). Comparison of three computer office workstations offering forearm support: impact on upper limb posture and muscle activation. Ergonomics, 49(2), 139-160.

Delleman, N. J., Haslegrave, C. M., and Chaffin, D. B. (Eds.). (2004). Working postures and movements. CRC Press.

Dempsey, P. G., McGorry, R. W., and Maynard, W. S. (2005). A survey of tools and methods used by certified professional ergonomists. Applied ergonomics, 36(4), 489-503.

DePalma, M. J., Ketchum, J. M., and Saullo, T. (2011). What is the source of chronic low back pain and does age play a role?. Pain Medicine, 12(2), 224-233.

Dhimitri, K. C., McGwin Jr, G., McNeal, S. F., Lee, P., Morse, P. A., Patterson, M., Wertz, F.D., and Marx, J. L. (2005). Symptoms of musculoskeletal disorders in ophthalmologists. American journal of ophthalmology, 139(1), 179-181.

Dickey, J. P., Eger, T. R., Frayne, R. J., Delgado, G. P., and Ji, X. (2013). Research Using Virtual Reality: Mobile Machinery Safety in the 21st Century. Minerals, 3(2), 145-164.

Douphrate, D. I., Fethke, N. B., Nonnenmann, M. W., Rosecrance, J. C., and Reynolds, S. J. (2012). Full shift arm inclinometry among dairy parlor workers: a feasibility study in a challenging work environment. Applied ergonomics, 43(3), 604-613.

Douphrate, D. I., Nonnenmann, M. W., and Rosecrance, J. C. (2009). Ergonomics in industrialized dairy operations. Journal of agromedicine, 14(4), 406-412.

Duc, C., Salvia, P., Lubansu, A., Feipel, V., and Aminian, K. (2014). A wearable inertial system to assess the cervical spine mobility: Comparison with an optoelectronic-based motion capture evaluation. Medical engineering and physics, 36(1), 49-56.

Dunning, K. K., Davis, K. G., Cook, C., Kotowski, S. E., Hamrick, C., Jewell, G., and Lockey, J. (2010). Costs by industry and diagnosis among musculoskeletal claims in a state workers compensation system: 1999–2004. American journal of industrial medicine, 53(3), 276-284.

Elford, W., Straker, L., and Strauss, G. (2000). Patient handling with and without slings: an analysis of the risk of injury to the lumbar spine. Applied ergonomics, 31(2), 185-200.

El-Gohary, M., and McNames, J. (2012). Shoulder and elbow joint angle tracking with inertial sensors. Biomedical Engineering, IEEE Transactions on, 59(9), 2635-2641.

Ettelt, D., Rey, P., Jourdan, G., Walther, A., Robert, P., and Delamare, J. (2014). 3D Magnetic Field Sensor Concept for Use in Inertial Measurement Units (IMUs).

Favre, J., Jolles, B. M., Aissaoui, R., and Aminian, K. (2008). Ambulatory measurement of 3D knee joint angle. Journal of biomechanics, 41(5), 1029–1035.

Favre, J., Jolles, B. M., Siegrist, O., and Aminian, K. (2006). Quaternion-based fusion of gyroscopes and accelerometers to improve 3D angle measurement. Electronics Letters, 42(11), 612-614.

Ferguson, S., Gaudes-MacLaren, L., Marras, W., Waters, T., and Davis, K. (2002). Spinal loading when lifting from industrial storage bins. Ergonomics, 45(6), 399-414.

Feng, Y., Grooten, W., Wretenberg, P., and Arborelius, U. P. (1997). Effects of arm support on shoulder and arm muscle activity during sedentary work. Ergonomics, 40(8), 834-848.

Ferrari, A., Cutti, A. G., Garofalo, P., Raggi, M., Heijboer, M., Cappello, A., and Davalli, A. (2009). First in vivo assessment of "Outwalk": a novel protocol for clinical gait analysis based on inertial and magnetic sensors. Medical and Biological Engineering and Computing, 48(1), 1–15.

Fethke, N. B., Gant, L. C., and Gerr, F. (2011). Comparison of biomechanical loading during use of conventional stud welding equipment and an alternate system. Applied ergonomics, 42(5), 725-734.

Fong, D. T. P., and Chan, Y. Y. (2010). The use of wearable inertial motion sensors in human lower limb biomechanics studies: A systematic review. Sensors, 10(12), 11556-11565.

Forsman, M., Hansson, G. Å., Medbo, L., Asterland, P., and Engström, T. (2002). A method for evaluation of manual work using synchronised video recordings and physiological measurements. Applied ergonomics, 33(6), 533-540.

Frames, C., Soangra, R., and Lockhart, T. E. (2013). Assessment of Postural Stability using Inertial Measurement Unit on Inclined Surfaces in Healthy Adults. Biomedical sciences instrumentation, 49, 234.

Gallagher, S., Marras, W. S., Davis, K. G., and Kovacs, K. (2002). Effects of posture on dynamic back loading during a cable lifting task. Ergonomics, 45(5), 380-398.

Gallagher, A., Matsuoka, Y., and Ang, W. T. (2004). An efficient real-time human posture tracking algorithm using low-cost inertial and magnetic sensors, in Proc. IEEE Int. Conf. Robot. Autom., Sendai, Japan, Sep. 28-Oct. 2, 2004, pp. 2967–2972.

Garbin, A. J. I., Garbin, C. A. S., Moimaz, S. A. S., Baldan, R. C. F., and Zina, L. G. (2011). Dental practice and musculoskeletal disorders association: A look at the evidence. Archives of Environmental and Occupational Health, 66(1), 26–33.

Garg, A., and Kapellusch, J. M. (2009). Applications of biomechanics for prevention of work-related musculoskeletal disorders. Ergonomics, 52(1), 36-59.

Genaidy, A. M., Al-Shedi, A. A., and Karwowski, W. (1994). Postural stress analysis in industry. Applied Ergonomics, 25(2), 77-87.

Gerr, F., Fethke, N., Anton, D., Merlino, L., Rosecrance, J., Marcus, M., and Jones, M. P. (2013). A Prospective Study of Musculoskeletal Outcomes Among Manufacturing Workers II. Effects of Psychosocial Stress and Work Organization Factors. Human Factors: The Journal of the Human Factors and Ergonomics Society.

Gerr, F., Fethke, N. B., Anton, D., Merlino, L., Rosecrance, J., Marcus, M., and Jones, M. P. (2014). A Prospective Study of Musculoskeletal Outcomes Among Manufacturing Workers II. Effects of Psychosocial Stress and Work Organization Factors. Human Factors: The Journal of the Human Factors and Ergonomics Society, 56(1), 178-190.

Giansanti, D., Maccioni, G., Benvenuti, F., and Macellari, V. (2007). Inertial measurement units furnish accurate trunk trajectory reconstruction of the sit-to-stand manoeuvre in healthy subjects. Medical and Biological Engineering and Computing, 45(10), 969–976.

Gill, K. P., and Callaghan, M. J. (1996). Intratester and intertester reproducibility of the lumbar motion monitor as a measure of range, velocity and acceleration of the thoracolumbar spine. Clinical Biomechanics, 11(7), 418-421.

Godwin, A., Agnew, M., and Stevenson, J. (2009). Accuracy of inertial motion sensors in static, quasistatic, and complex dynamic motion. Journal of biomechanical engineering, 131(11), 114501.

Gofrit, O. N., Mikahail, A. A., Zorn, K. C., Zagaja, G. P., Steinberg, G. D., and Shalhav, A. L. (2008). Surgeons' perceptions and injuries during and after urologic laparoscopic surgery. Urology, 71(3), 404-407.

Goodvin, C., Park, E. J., Huang, K., and Sakaki, K. (2006). Development of a real-time three-dimensional spinal motion measurement system for clinical practice. Medical and Biological Engineering and Computing, 44(12), 1061–1075.

Gouwanda, D., and Senanayake, S. (2008). Emerging trends of body-mounted sensors in sports and human gait analysis. International Conference on Biomedical Engineering, 715-718.

Greene, B. R., McGrath, D., O'Neill, R., O'Donovan, K. J., Burns, A., and Caulfield, B. (2010). An adaptive gyroscope-based algorithm for temporal gait analysis. Medical and biological engineering and computing, 48(12), 1251-1260.

Haddad, O., Sanjari, M. A., Amirfazli, A., Narimani, R., and Parnianpour, M. (2012). Trapezius muscle activity in using ordinary and ergonomically designed dentistry chairs. International Journal of Occupational and Environmental Medicine, 3(2).

Hagberg, M., and Wegman, D. H. (1987). Prevalence rates and odds ratios of shoulder-neck diseases in different occupational groups. British journal of industrial medicine, 44(9), 602-610.

Hägg, G. M., €Oster, J., and Bystr€om, S. (1997). Forearm muscular load and wrist angle among automobile assembly line workers in relation to symptoms. Applied Ergonomics, 28(1), 41–47.

Hansson, G. Å., Arvidsson, I., Ohlsson, K., Nordander, C., Mathiassen, S. E., Skerfving, S., and Balogh, I. (2006). Precision of measurements of physical workload during standardised manual handling. Part II: Inclinometry of head, upper back, neck and upper arms. Journal of Electromyography and Kinesiology, 16(2), 125-136.

Hansson, G. Å., Asterland, P., Holmer, N. G., and Skerfving, S. (2001a). Validity and reliability of triaxial accelerometers for inclinometry in posture analysis. Medical and Biological Engineering and Computing, 39(4), 405-413.

Hansson, G. Å., Balogh, I., Byström, J. U., Ohlsson, K., Nordander, C., Asterland, P., Sjölander, S., Rylander, L., Winkel, J., and Skerfving, S. (2001b). Questionnarie versus direct technical measurements in assessing postures and movements of the head, upper back, arms and hands. Scandinavian journal of work, environment and health, 30-40.

Hansson, G. Å., Balogh, I., Ohlsson, K., and Skerfving, S. (2004). Measurements of wrist and forearm positions and movements: effect of, and compensation for, goniometer crosstalk. Journal of Electromyography and Kinesiology, 14(3), 355–367.

Hansson, G. Å., Balogh, I., Ohlsson, K., Granqvist, L., Nordander, C., Arvidsson, I., Akesson, I., Unge, J., Rittner, R., Strömberg, U., and Skerfving, S. (2010). Physical workload in various types of work: Part II. Neck, shoulder and upper arm. International Journal of Industrial Ergonomics, 40(3), 267-281.

Hayes, M. J., Cockrell, D., and Smith, D. R. (2009). A systematic review of musculoskeletal disorders among dental professionals. International journal of dental hygiene, 7(3), 159-165.

Hayes, M. J., Smith, D. R., and Taylor, J. A. (2013). Musculoskeletal disorders and symptom severity among Australian dental hygienists. BMC Research Notes, 6(1), 250.

Hengel, K. M. O., Visser, B., and Sluiter, J. K. (2011). The prevalence and incidence of musculoskeletal symptoms among hospital physicians:A systematic review. International Archives of Occupational and Environmental Health, 84(2), 115–119.

Henriksen, M., Lund, H., Bliddal, H., and Danneskiold-Samsøe, B. (2007). Dynamic control of the lumbopelvic complex; lack of reliability of established test procedures. European Spine Journal, 16(6), 733-740.

Higgins, W. T. (1975). A comparison of complementary and Kalman filtering. IEEE Transactions on Aerospace and Electronic Systems, 3, 321-325.

Hignett, S., and McAtamney, L. (2000). Rapid Entire Body Assessment (REBA). Applied Ergonomics. 31, 201-205.

Hooftman, W. E., van der Beek, A. J., van de Wal, B. G., Knol, D. L., Bongers, P. M., Burdorf, A., and van Mechelen, W. (2009). Equal task, equal exposure? Are men and women with the same tasks equally exposed to awkward working postures?. Ergonomics, 52(9), 1079-1086.

Hoogendoorn, W. E., Bongers, P. M., de Vet, H. C., Douwes, M., Koes, B. W., Miedema, M. C., Ariens, G.A., and Bouter, L. M. (2000). Flexion and rotation of the trunk and lifting at work are risk factors for low back pain: results of a prospective cohort study. Spine, 25(23), 3087-3092.

Hoy, D., Bain, C., Williams, G., March, L., Brooks, P., Blyth, F., Woolf, A., Vos, T., and Buchbinder, R. (2012). A systematic review of the global prevalence of low back pain. Arthritis and Rheumatism, 64(6), 2028-2037.

Jansen, J. P., Morgenstern, H., and Burdorf, A. (2004). Dose-response relations between occupational exposures to physical and psychosocial factors and the risk of low back pain. Occupational and Environmental Medicine, 61(12), 972-979.

Jasiewicz, J. M., Treleaven, J., Condie, P., and Jull, G. (2007). Wireless orientation sensors: their suitability to measure head movement for neck pain assessment. Manual therapy, 12(4), 380-385.

Jiang, C., Xue, L., Chang, H., Yuan, G., and Yuan, W. (2012). Signal Processing of MEMS Gyroscope Arrays to Improve Accuracy Using a 1st Order Markov for Rate Signal Modeling. Sensors, 12(2), 1720–1737.

Jonsson, B. (1988). The static load component in muscle work. European journal of applied physiology and occupational physiology, 57(3), 305-310.

Juul-Kristensen, B., Hansson, G. Å., Fallentin, N., Andersen, J. H., and Ekdahl, C. (2001). Assessment of work postures and movements using a video-based observation method and direct technical measurements. Applied ergonomics, 32(5), 517-524.

Kalman, R. E. (1960). A new approach to linear filtering and prediction problems. Journal of basic Engineering, 82(1), 35-45.

Karhu, O., Kansi, P., and Kuorinka, I. (1977). Correcting working postures in industry: a practical method for analysis. Applied Ergonomics. 8, 199-201.

Karpovich, P. V., and Wilklow, L. B. (1959). A Goniometric Study of the Human Foot in Standing and Walking. U.S. Armed Forces Medical Journal, 10, 885-903.

Karpovich, P. V., Herden Jr, E. L., and Asa, M. M. (1960). Electrogoniometric study of joints. United States Armed Forces Medical Journal, 11, 424.

Kazmierczak, K., Mathiassen, S. E., Forsman, M., and Winkel, J. (2005). An integrated analysis of ergonomics and time consumption in Swedish ‘craft-type’car disassembly. Applied Ergonomics, 36(3), 263-273.

Kazmierczak, K., Mathiassen, S. E., Neumann, P., and Winkel, J. (2006). Observer reliability of industrial activity analysis based on video recordings. International journal of industrial ergonomics, 36(3), 275-282.

Kerr, M. S., Frank, J. W., Shannon, H. S., Norman, R. W., Wells, R. P., Neumann, W. P., Bombardier, C., and Ontario Universities Back Pain Study Group. (2001). Biomechanical and psychosocial risk factors for low back pain at work. American journal of public health, 91(7), 1069.

Keyserling, W. M. (1986). Postural analysis of the trunk and shoulders in simulated real time. Ergonomics, 29(4), 569-583.

Kim, S., and Nussbaum, M. A. (2013). Performance evaluation of a wearable inertial motion capture system for capturing physical exposures during manual material handling tasks. Ergonomics, 56(2), 314-326.

Kitzmann, A. S., Fethke, N. B., Baratz, K. H., Zimmerman, M. B., Hackbarth, D. J., and Gehrs, K. M. (2012). A survey study of musculoskeletal disorders among eye care physicians compared with family medicine physicians. Ophthalmology, 119(2), 213-220.

Koehoorn, M. (2010). Evaluating Methods to Use the Virtual Corset™ Inclinometer for Trunk Posture and Spinal Compression Measurements. Final Report to WorkSafeBC.

Leboeuf-Yde, C., Lauritsen, J. M., and Lauritzen, T. (1997). Why has the search for causes of low back pain largely been nonconclusive?. Spine, 22(8), 877-881.

Leclerc, A., Chastang, J. F., Niedhammer, I., Landre, M. F., and Roquelaure, Y. (2004). Incidence of shoulder pain in repetitive work. Occupational and environmental medicine, 61(1), 39-44.

Lee, J. K., and Park, E. J. (2011). Quasi real-time gait event detection using shank-attached gyroscopes. Medical and biological engineering and computing, 49(6), 707-712.

Lee, R. Y., Laprade, J., and Fung, E. H. (2003). A real-time gyroscopic system for three-dimensional measurement of lumbar spine motion. Medical engineering and physics, 25(10), 817-824.

Leigh, J., Macaskill, P., Kuosma, E., and Mandryk, J. (1999). Global burden of disease and injury due to occupational factors. Epidemiology. 10, 626-31.

Levitz, C.L. and Iannotti, J.P. (1995). Overuse injuries of the shoulder. In: Gordon SL, Blair SJ, Fine LJ, eds. Repetitive motions disorders of the upper extremity. Rosemont, IL: American Academy of Orthopaedic Surgeon.

Li, G., and Buckle, P. (1999). Current techniques for assessing physical exposure to work-related musculoskeletal risks, with emphasis on posture-based methods. Ergonomics, 42(5), 674-695.

Ligorio, G., and Sabatini, A. M. (2013). Extended Kalman filter-based methods for pose estimation using visual, inertial and magnetic sensors: comparative analysis and performance evaluation. Sensors, 13(2), 1919-1941.

Long, J., Burgess-Limerick, R., and Stapleton, F. (2014). Personal consequences of work-related physical discomfort: An exploratory study. Clinical and Experimental Optometry, 97(1), 30–35.

Long, J., Naduvilath, T. J., Hao, L. E., Li, A., Ng, W., Yip, W., and Stapleton, F. (2011). Risk factors for physical discomfort in Australian optometrists. Optometry and Vision Science, 88(2), 317-326.

Ludewig, P. M., and Cook, T. M. (2000). Alterations in shoulder kinematics and associated muscle activity in people with symptoms of shoulder impingement. Physical therapy, 80(3), 276-291.

Luime, J. J., Koes, B. W., Hendriksen, I. J. M., Burdorf, A., Verhagen, A. P., Miedema, H. S., and Verhaar, J. A. N. (2004). Prevalence and incidence of shoulder pain in the general population; a systematic review. Scandinavian journal of rheumatology, 33(2), 73-81.

Luinge, H. J., and Veltink, P. H. (2004). Inclination measurement of human movement using a 3-D accelerometer with autocalibration. Neural Systems and Rehabilitation Engineering, IEEE Transactions on, 12(1), 112-121.

Luinge, H. J., and Veltink, P. H. (2005). Measuring orientation of human body segments using miniature gyroscopes and accelerometers. Medical and Biological Engineering and computing, 43(2), 273-282.

Luinge, H. J., Veltink, P. H., and Baten, C. T. (2007). Ambulatory measurement of arm orientation. Journal of biomechanics, 40(1), 78-85.

Luo, X., Pietrobon, R., S, X. S., Liu, G. G., and Hey, L. (2004). Estimates and patterns of direct health care expenditures among individuals with back pain in the United States. Spine, 29, 79–86.

Lyons, J. (2002). Factors contributing to low back pain among professional drivers: a review of current literature and possible ergonomic controls. Work: A Journal of Prevention, Assessment and Rehabilitation, 19(1), 95-102.

Manchikanti, L. (2000). Epidemiology of low back pain. Pain physician, 3(2), 167-192.

Marras, W. S. (2012). The Complex Spine The Multidimensional System of Causal Pathways for Low-Back Disorders. Human Factors: The Journal of the Human Factors and Ergonomics Society, 54(6), 881-889.

Marras W. S., and W. G. Allread. (2006). Lumbar Motion Monitor. In N. A. Stanton, A. Hedge, K. Brookhuis, E. Salas, and H. Hendrick. (Eds.), Handbook of Human Factors methods. Boca Raton, USA, CRC Press.

Marras, W. S., Cutlip, R. G., Burt, S. E., and Waters, T. R. (2009). National occupational research agenda (NORA) future directions in occupational musculoskeletal disorder health research. Applied Ergonomics, 40(1), 15–22.

Marras, W. S., Fathallah, F. A., Miller, R. J., Davis, S. W., and Mirka, G. A. (1992). Accuracy of a three-dimensional lumbar motion monitor for recording dynamic trunk motion characteristics. International Journal of Industrial Ergonomics, 9(1), 75-87.

Marras, W. S., Lavender, S. A., Leurgans, S. E., Fathallah, F.A., Ferguson, S.A., Allread, W.G., and Rajulu, S.L. (1995). Biomechanical risk factors for occupationally related low back disorders. Ergonomics, 38(2), 377-410.

Marras, W. S., Lavender, S., Leurgans, S., Rajulu, S., Allread, W., Fathallah, F., Ferguson, S. (1993). The role of dynamic three-dimensional trunk motion in occupationally-related low back disorders. Spine, 18 (5), 617-628.

Marx, J. L. (2012). Ergonomics: Back to the Future. Ophthalmology, 119(2), 211-212.

Marx, J.L., Wertz F.D., Dhimitri, K.C. (2005). Work-related musculoskeletal disorders in ophthalmologists. Techniques in Ophthalmology, 3 (1), 54-61.

Matern, U. (2009). Ergonomic deficiencies in the operating room: Examples from minimally invasive surgery. Work: A Journal of Prevention, Assessment and Rehabilitation, 33(2), 165–168.

Mathiassen, S. E. (2006). Diversity and variation in biomechanical exposure: what is it, and why would we like to know?. Applied ergonomics, 37(4), 419-427.

Mathiassen, S. E., Wahlström, J., and Forsman, M. (2012). Bias and imprecision in posture percentile variables estimated from short exposure samples. BMC medical research methodology, 12(1), 36.

Mathiassen, S. E., Nordander, C., Svendsen, S. W., Wellman, H. M., and Dempsey, P. G. (2005). Task-based estimation of mechanical job exposure in occupational groups. Scandinavian journal of work, environment and health, 31(2), 138-151.

Mathiassen, S. E., Winkel, J., and Hägg, G. M. (1995). Normalization of surface EMG amplitude from the upper trapezius muscle in ergonomic studies—a review. Journal of electromyography and kinesiology, 5(4), 197-226.

McCauley-Bush, P. (2011). Ergonomics: Foundational Principles, Applications, and Technologies. CRC Press, Taylor and Francis, Boca Raton, FL.

Meislin, R. J., Sperling, J. W., and Stitik, T. P. (2005). Persistent shoulder pain: epidemiology, pathophysiology, and diagnosis. American journal of orthopedics (Belle Mead, NJ), 34(12 Suppl), 5-9.

Mieritz, R. M., Bronfort, G., Kawchuk, G., Breen, A., and Hartvigsen, J. (2012). Reliability and measurement error of 3-dimensional regional lumbar motion measures: a systematic review. Journal of manipulative and physiological therapeutics, 35(8), 645-656.

Milerad, E., and Ericson, M. O. (1994). Effects of precision and force demands, grip diameter, and arm support during manual work: an electromyographic study. Ergonomics, 37(2), 255-264.

Miller, C., Mulavara, A., and Bloomberg, J. (2002). A quasi-static method for determining the characteristics of a motion capture camera system in a "split-volume" configuration. Gait and posture, 16(3), 283-287.

Miranda, H., Viikari-Juntura, E., Heistaro, S., Heliövaara, M., and Riihimäki, H. (2005). A population study on differences in the determinants of a specific shoulder disorder versus nonspecific shoulder pain without clinical findings. American journal of epidemiology, 161(9), 847-855.

Miranda, H., Viikari-Juntura, E., Martikainen, R., Takala, E. P., and Riihimäki, H. (2001). A prospective study of work related factors and physical exercise as predictors of shoulder pain. Occupational and Environmental Medicine, 58(8), 528-534.

Moore, J.S., and Garg, A. (1995). The Strain Index: a proposed method to analyze jobs for risk of distal upper extremity disorders. American Industrial Hygiene Association Journal, 55, 703-715.

Morse, T., Dillon, C., Kenta-Bibi, E., Weber, J., Diva, U., Warren, N., and Grey, M. (2005). Trends in work-related musculoskeletal disorder reports by year, type, and industrial sector: A capture-recapture analysis. American Journal of Industrial Medicine, 48(1), 40–49.

Murphy, R. J., and Carr, A. J. (2010). Shoulder pain. Clin Evid (Online), 2010.

Murray, C. J., Vos, T., Lozano, R., Naghavi, M., Flaxman, A. D., Michaud, C., et al. (2013). Disability-adjusted life years (DALYs) for 291 diseases and injuries in 21 regions, 1990–2010: a systematic analysis for the Global Burden of Disease Study 2010. The Lancet, 380(9859), 2197-2223.

National Institute for Occupational Safety and Health. (1996). National occupational research agenda. Cincinnati, OH: U.S. Department of Health and Human Services, Public Health Service, Centers for Disease Control and Prevention, National Institute for Occupational Safety and Health, DHHS (NIOSH) Publication No. 96–115. URL http://www.cdc.gov/niosh/docs/96-115/default.html

National Institute for Occupational Safety and Health. (1997). Musculoskeletal Disorders and Workplace Factors: a critical review of epidemiologic evidence for work-related musculoskeletal disorders of the neck, upper extremity, and lower back. Bernard, B. P, Ed., DHHS (NIOSH) Publication No. 97-141. URL http://www.cdc.gov/niosh/docs/97-141/.

National Institute for Occupational Safety and Health. (2001). National Occupational Research Agenda for Musculoskeletal Disorders: Research Topics for the Next Decade, A Report by the NORA Musculoskeletal Disorders Team. DHHS (NIOSH) Publication No. 2001-117. URL http://www.cdc.gov/niosh/2001-117.html

National Institute for Occupational Safety and Health. (2014). Observation-based posture assessment: review of current practice and recommendations for improvement. By Lowe BD, Weir PL, Andrews DM. Cincinnati, OH: U.S. Department of Health and Human Services, Centers for Disease Control and Prevention, National Institute for Occupational Safety and Health, DHHS (NIOSH) Publication No. 2014–131.

National Research Council and the Institute of Medicine (2001). Musculoskeletal disorders and the workplace: low back and upper extremities. Panel on Musculoskeletal Disorders and the Workplace. Commission on Behavioral and Social Sciences and Education. Washington, DC: National Academy Press. URL http://www.nap.edu/openbook.php?isbn=0309072840

Nieminen, H., Takala, E. P., and Viikari-Juntura, E. (1993). Normalization of electromyogram in the neck-shoulder region. European Journal of Applied Physiology and Occupational Physiology, 67(3), 199–207.

Ngan, K., Drebit, S., Siow, S., Yu, S., Keen, D., and Alamgir, H. (2010). Risks and causes of musculoskeletal injuries among health care workers. Occupational medicine, 60(5), 389-394.

Nimbarte, A. D., Sivak-Callcott, J. A., Zreiqat, M., and Chapman, M. (2013). Neck postures and cervical spine loading among microsurgeons operating with loupes and headlamp. IIE Transactions on Occupational Ergonomics and Human Factors, 1(4), 215–223.

Nirschl, R. P. (1988). Rotator cuff tendinitis: basic concepts of pathoetiology. Instructional course lectures, 38, 439-445.

Occhionero, V., Korpinen, L., and Gobba, F. (2014). Upper limb musculoskeletal disorders in healthcare personnel. Ergonomics, (ahead-of-print), 1-26.

Odell, D., Barr, A., Goldberg, R., Chung, J., and Rempel, D. (2007). Evaluation of a dynamic arm support for seated and standing tasks: A laboratory study of electromyography and subjective feedback. Ergonomics, 50(4), 520–535.

O’Donovan, K. J., Bourke, A. K., O’Keeffe, D. T., and ÓLaighin, G. (2009). The application of inertial and magnetic sensors to the monitoring of calf muscle pump activity. Medical engineering and physics, 31(1), 55-60.

Ohlsson, K., Attewell, R. G., Johnsson, B., Ahlm, A., and Skerfving, S. (1994). An assessment of neck and upper extremity disorders by questionnaire and clinical examination. Ergonomics, 37(5), 891-897.

Osborn, J., Newell, K., Rudney, J., and Stoltenberg, J. (1990). Musculoskeletal pain among Minnesota dental hygienists. Journal of Dental Hygiene: JDH/American Dental Hygienists' Association, 64(3), 132–138.

O'Sullivan, K., O'Sullivan, L., Campbell, A., O'Sullivan, P., and Dankaerts, W. (2012). Towards monitoring lumbo-pelvic posture in real-life situations: Concurrent validity of a novel posture monitor and a traditional laboratory-based motion analysis system. Man Ther, 17(1), 77-83.

Palmer, K. T., Harris, E. C., Linaker, C., Cooper, C., and Coggon, D. (2012). Optimising case definitions of upper limb disorder for aetiological research and prevention: a review. Occupational and environmental medicine, 69(1), 71-78.

Paquet, V. L., Punnett, L., and Buchholz, B. (2001). Validity of fixed-interval observations for postural assessment in construction work. Applied ergonomics, 32(3), 215-224.

Pelham, T. W., White, H., and, L. E. H., and Lee, S. W. (2005). The etiology of low back pain in military helicopter aviators: prevention and treatment. Work: A Journal of Prevention, Assessment and Rehabilitation, 24(2), 101-110.

Peng, B., Wu, W., Hou, S., Li, P., Zhang, C., and Yang, Y. (2005). The pathogenesis of discogenic low back pain. Journal of Bone and Joint Surgery, British Volume, 87(1), 62-67.

Picerno, P., Cereatti, A., and Cappozzo, A. (2008). Joint kinematics estimate using wearable inertial and magnetic sensing modules. Gait and posture, 28(4), 588-595.

Piedrahíta, H. (2006). Costs of work-related musculoskeletal disorders (MSDs) in developing countries: Colombia case international. International Journal of Occupational Safety and Ergonomics, 12, 379-386.

Plamondon, A., Delisle, A., Larue, C., Brouillette, D., McFadden, D., Desjardins, P., and Larivière, C. (2007). Evaluation of a hybrid system for three-dimensional measurement of trunk posture in motion. Applied Ergonomics, 38(6), 697-712.

Pope, M. H., Goh, K. L., and Magnusson, M. L. (2002). Spine ergonomics. Annual review of biomedical engineering, 4(1), 49-68.

Praemer, A., Furner, S., and Rice, D. P. (1999). Musculoskeletal Conditions in the United States. Rosemont, Illinois: American Academy of Orthopaedic Surgeons.

Punnett, L. (2014). Musculoskeletal disorders and occupational exposures: How should we judge the evidence concerning the causal association?. Scandinavian journal of public health, 42(13 suppl), 49-58.

Punnett, L., Fine, L. J., Keyserling, W. M., Herrin, G. D., and Chaffin, D. B. (2000). Shoulder disorders and postural stress in automobile assembly work. Scandinavian journal of work, environment and health, 283-291.

Punnett, L., Fine, L. J., Keyserling, W. M., Herrin, G. D., and Chaffin, D. B. (1991). Back disorders and nonneutral trunk postures of automobile assembly workers. Scandinavian journal of work, environment and health, 337-346.

Punnett, L., and Wegman, D. H. (2004). Work-related musculoskeletal disorders: the epidemiologic evidence and the debate. Journal of Electromyography and Kinesiology, 14(1), 13-23.

Rab, G., Petuskey, K., and Bagley, A. (2002). A method for determination of upper extremity kinematics. Gait and posture, 15(2), 113-119.

Rahmatalla, S., Xia, T., Contratto, M., Wilder, D., Frey-Law, L., Kopp, G., and Grosland, N. (2006). 3D Displacement, Velocity, and Acceleration of Seated Operators in Whole-Body Vibration Environment using Optical Motion Capture Systems. In The Ninth International Symposium on the 3-D Analysis of Human Movement, Valenciennes, France, 28-30 June, 2006.

Rezagholi, M., Mathiassen, S. E., and Liv, P. (2012). Cost efficiency comparison of four video-based techniques for assessing upper arm postures. Ergonomics, 55(3), 350-360.

Ribeiro, D.C., Sole, G., Abbott, J.H. and Milosavljevic, S., 2011. Extrinsic feedback and management of low back pain: A critical review of the literature. Manual therapy, 16 (3), 231-239.

Ribeiro, D.C., Sole, G., Abbott, J.H. and Milosavljevic, S., 2013. Validity and reliability of the spineangel® lumbo-pelvic postural monitor. Ergonomics, 56 (6), 977-991.

Ribeiro, D.C., Sole, G., Abbott, J.H. and Milosavljevic, S., 2014. The effectiveness of a lumbopelvic monitor and feedback device to change postural behavior: A feasibility randomized controlled trial. journal of orthopaedic & sports physical therapy, 44 (9), 702-711.

Rota, S., Rogowski, I., Champely, S., and Hautier, C. (2013). Reliability of EMG normalisation methods for upper-limb muscles. Journal of Sports Sciences, 31(15), 1696–1704.

Roetenberg, D., Luinge, H. J., Baten, C. T., and Veltink, P. H. (2005). Compensation of magnetic disturbances improves inertial and magnetic sensing of human body segment orientation. Neural Systems and Rehabilitation Engineering, IEEE Transactions on, 13(3), 395-405.

Roetenberg, D., Slycke, P. J., and Veltink, P. H. (2007). Ambulatory position and orientation tracking fusing magnetic and inertial sensing. Biomedical Engineering, IEEE Transactions on, 54(5), 883-890.

Roffey, D. M., Wai, E. K., Bishop, P., Kwon, B. K., and Dagenais, S. (2010). Causal assessment of awkward occupational postures and low back pain: results of a systematic review. The Spine Journal, 10(1), 89-99.

Rota, S., Rogowski, I., Champely, S., and Hautier, C. (2013). Reliability of EMG normalisation methods for upper-limb muscles. Journal of sports sciences, (ahead-of-print), 1-9.

Rubin, D. I. (2007). Epidemiology and risk factors for spine pain. Neurologic clinics, 25(2), 353-371.

Sabatini, A. M. (2011). Kalman-filter-based orientation determination using inertial/magnetic sensors: Observability analysis and performance evaluation. Sensors, 11(10), 9182-9206.

Sabatini, A. M. (2006). Quaternion-based extended Kalman filter for determining orientation by inertial and magnetic sensing. Biomedical Engineering, IEEE Transactions on, 53(7), 1346-1356.

Schepers, H. M., Roetenberg, D., and Veltink, P. H. (2010). Ambulatory human motion tracking by fusion of inertial and magnetic sensing with adaptive actuation. Medical and Biological Engineering and Computing, 48(1), 27–37.

Seidler, A., Bolm-Audorff, U., Heiskel, H., Henkel, N., Roth-Küver, B., Kaiser, U., Bickeboller, W.J., Beck, W., and Elsner, G. (2001). The role of cumulative physical work load in lumbar spine disease: risk factors for lumbar osteochondrosis and spondylosis associated with chronic complaints. Occupational and environmental medicine, 58(11), 735-746.

Silverstein, B. A., Bao, S. S., Fan, Z. J., Howard, N., Smith, C., Spielholz, P., Bonauto, D., and Viikari-Juntura, E. (2008). Rotator cuff syndrome: personal, work-related psychosocial and physical load factors. Journal of Occupational and Environmental Medicine, 50(9), 1062-1076.

Silverstein, B. A., Viikari-Juntura, E., Fan, Z. J., Bonauto, D. K., Bao, S., and Smith, C. (2006). Natural course of nontraumatic rotator cuff tendinitis and shoulder symptoms in a working population. Scandinavian journal of work, environment and health, 32(2), 99-108.

Sivak-Callcott, J. A., Diaz, S. R., Ducatman, A. M., Rosen, C. L., Nimbarte, A. D., and Sedgeman, J. A. (2011). A survey study of occupational pain and injury in ophthalmic plastic surgeons. Ophthalmic Plastic and Reconstructive Surgery, 27(1), 28–32.

Solomonow, M., Baratta, R. V., Banks, A., Freudenberger, C., and Zhou, B. H. (2003). Flexion–relaxation response to static lumbar flexion in males and females. Clinical Biomechanics, 18(4), 273-279.

Srinivasan, D., and Mathiassen, S. E. (2012). Motor variability–an important issue in occupational life. Work: A Journal of Prevention, Assessment and Rehabilitation, 41, 2527-2534.

Srinivasan, D., Mathiassen, S. E., Samani, A., and Madeleine, P. (2014). The combined influence of task accuracy and pace on motor variability in a standardized repetitive precision task. Ergonomics.

Stomberg, M. W., Tronstad, S. E., Hedberg, K., Bengtsson, J., Jonsson, P., Johansen, L., and Lindvall, B. (2010). Work-related musculoskeletal disorders when performing laparoscopic surgery. Surgical Laparoscopy Endoscopy and Percutaneous Techniques, 20(1), 49-53.

Svendsen, S. W., Bonde, J. P., Mathiassen, S. E., Stengaard-Pedersen, K., and Frich, L. H. (2004a). Work related shoulder disorders: quantitative exposure-response relations with reference to arm posture. Occupational and environmental medicine, 61(10), 844-853.

Svendsen, S. W., Gelineck, J., Mathiassen, S. E., Bonde, J. P., Frich, L. H., Stengaard-Pedersen, K., and Egund, N. (2004b). Work above shoulder level and degenerative alterations of the rotator cuff tendons: a magnetic resonance imaging study. Arthritis and Rheumatism, 50(10), 3314-3322.

Szeto, G. P., Ho, P., Ting, A. C., Poon, J. T., Cheng, S. W., and Tsang, R. C. (2009). Work-related musculoskeletal symptoms in surgeons. Journal of occupational rehabilitation, 19(2), 175-184.

Takala, E. P., Pehkonen, I., Forsman, M., Hansson, G. Å., Mathiassen, S. E., Neumann, W. P., Sjøgaard, G., Veiersted, K.B., Westgaard, R.H., and Winkel, J. (2010). Systematic evaluation of observational methods assessing biomechanical exposures at work. Scandinavian journal of work, environment and health, 3-24.

Tanaka, S., Petersen, M., and Cameron, L. (2001). Prevalence and risk factors of tendinitis and related disorders of the distal upper extremity among US workers:

Comparison to carpal tunnel syndrome*†. American journal of industrial medicine, 39(3), 328-335.

Teschke, K., Trask, C., Johnson, P., Chow, Y., Village, J., and Koehoorn, M. (2009). Measuring posture for epidemiology: comparing inclinometry, observations and self-reports. Ergonomics, 52(9), 1067-1078.

Theobald, P. S., Jones, M. D., and Williams, J. M. (2012). Do inertial sensors represent a viable method to reliably measure cervical spine range of motion?. Manual therapy, 17(1), 92-96.

Thorn, S., Søgaard, K., Kallenberg, L. A. C., Sandsjö, L., Sjøgaard, G., Hermens, H. J., Kadefors, R., and Forsman, M. (2007). Trapezius muscle rest time during standardised computer work–a comparison of female computer users with and without self-reported neck/shoulder complaints. Journal of Electromyography and Kinesiology, 17(4), 420-427.

Trask, C. M., Teschke, K., Morrison, J., Johnson, P. W., Village, J., and Koehoorn, M. (2008). How long is long enough? Evaluating sampling durations for low back EMG assessment. Journal of occupational and environmental hygiene, 5(10), 664-670.

Trask, C., Teschke, K., Morrison, J., Village, J., Johnson, P., and Koehoorn, M. (2010). Using observation and self-report to predict mean, 90th percentile, and cumulative low back muscle activity in heavy industry workers. Annals of occupational hygiene, 54(5), 595-606.

Trask, C., Teschke, K., Village, J., Chow, Y., Johnson, P., Luong, N., and Koehoorn, M. (2007). Measuring low back injury risk factors in challenging work environments: an evaluation of cost and feasibility. American journal of industrial medicine, 50(9), 687-696.

Trejo, A., Jung, M. C., Oleynikov, D., and Hallbeck, M. S. (2007). Effect of handle design and target location on insertion and aim with a laparoscopic surgical tool. Applied Ergonomics, 38(6), 745–753.

Van der Schatte Olivier, R., van't Hullenaar, C., Ruurda, J., and Broeders, I. (2009). Ergonomics, user comfort, and performance in standard and robot-assisted laparoscopic surgery. Surgical Endoscopy, 23(6), 1365–1371.

van der Windt, D. A., Thomas, E., Pope, D. P., de Winter, A. F., Macfarlane, G. J., Bouter, L. M., and Silman, A. J. (2000). Occupational risk factors for shoulder pain: a systematic review. Occupational and environmental medicine, 57(7), 433-442.

Van Driel, R., Teschke, K., Callaghan, J.P., Trask, C., Koehoorn, M., and Johnson, P.W. (2009). A Comparison of Trunk Posture Movements: A Motion Capture System and a New Data-Logging Inclinometer. Presented at IEA 2009, 17thWorld Conference on Ergonomics. Beijing, China.

Van Driel, R., Trask, C., Johnson, P. W., Callaghan, J. P., Koehoorn, M., and Teschke, K. (2013). Anthropometry-Corrected Exposure Modeling as a Method to Improve Trunk Posture Assessment with a Single Inclinometer. Journal of occupational and environmental hygiene, 10(3), 143-154.

van Oostrom, S. H., Verschuren, M., de Vet, H. C., Boshuizen, H. C., and Picavet, H. S. J. (2012). Longitudinal associations between physical load and chronic low back pain in the general population: The doetinchem cohort study. Spine, 37(9), 788-796.

van Veelen, M., Jakimowicz, J., and Kazemier, G. (2004). Improved physical ergonomics of laparoscopic surgery. Minimally Invasive Therapy and Allied Technologies, 13(3), 161–166.

Verstraete, M. C., and Soutas-Little, R. W. (1990). A method for computing the three-dimensional angular velocity and acceleration of a body segment from three-dimensional position data. Journal of biomechanical engineering, 112(2), 114-118.

Vieira, E. R., and Kumar, S. (2004). Working postures: a literature review. Journal of Occupational Rehabilitation, 14(2), 143-159.

Viikari-Juntura, E., Martikainen, R., Luukkonen, R., Mutanen, P., Takala, E. P., and Riihimäki, H. (2001). Longitudinal study on work related and individual risk factors affecting radiating neck pain. Occupational and environmental medicine, 58(5), 345-352.

Wagenaar, R. C., Sapir, I., Zhang, Y., Markovic, S., Vaina, L. M., and Little, T. D. (2011). Continuous monitoring of functional activities using wearable, wireless gyroscope and accelerometer technology. In Engineering in Medicine and Biology Society, EMBC, 2011 Annual International Conference of the IEEE (pp. 4844-4847). IEEE.

Wahlstrom, J., Mathiassen, S. E., Liv, P., Hedlund, P., Ahlgren, C., and Forsman, M. (2010). Upper arm postures and movements in female hairdressers across four full working days. Annals of occupational hygiene, 54(5), 584-594.

Wai, E. K., Roffey, D. M., Bishop, P., Kwon, B. K., and Dagenais, S. (2010a). Causal assessment of occupational bending or twisting and low back pain: results of a systematic review. The Spine Journal, 10(1), 76-88.

Wai, E. K., Roffey, D. M., Bishop, P., Kwon, B. K., and Dagenais, S. (2010b). Causal assessment of occupational lifting and low back pain: results of a systematic review. The Spine Journal, 10(6), 554–566.

Walker, B. F. (2000). The prevalence of low back pain: a systematic review of the literature from 1966 to 1998. Journal of Spinal Disorders and Techniques, 13(3), 205-217.

Walker, J. (2014). Shoulder pain: pathogenesis, diagnosis and management. Nursing Standard, 28(22), 51-58.

Wang, L., Hu, W., and Tan, T. (2003). Recent developments in human motion analysis. Pattern Recognition, 36, 585–601.

Waters, T. R., Putz-Anderson, V., Garg, A., and Fine, L. J. (1993). Revised NIOSH equation for the design and evaluation of manual lifting tasks. Ergonomics, 36(7), 749-776.

Waters, T., Collins, J., Galinsky, T., and Caruso, C. (2006). NIOSH research efforts to prevent musculoskeletal disorders in the healthcare industry. Orthopaedic Nursing, 25(6), 380-389.

Winkel, J., and Mathiassen, S.E. (1994). Assessment of physical work load in epidemiologic studies: concepts, issues and operational considerations. Ergonomics, 37, 979–988.

Winkel, J., and Westgaard, R. (2000). Occupational and individual risk factors for shoulder-neck complaints: Part II—The scientific basis (literature review) for the guide. Elsevier Ergonomics Book Series, 1, 83-102.

Wong, K. C., Lee, R. Y., and S. S. Yeung. (2009). “The association between back pain and trunk posture of workers in a special school for the severe handicaps.” BMC musculoskeletal disorders 10(1): 43.

Wong, W. Y., and Wong, M. S. (2008). Trunk posture monitoring with inertial sensors. European Spine Journal, 17(5), 743-753.

Woolf, A. D., and Pfleger, B. (2003). Burden of major musculoskeletal conditions. Bulletin of the World Health Organization, 81(9), 646-656.

Woolley, S.M., Kitzmann, A.S. (2011). The Importance of Ergonomics for Ophthalmologists. In: Oetting TA, ed. Basic Principles of Ophthalmic Surgery, 2nd ed. San Francisco: American Academy of Ophthalmology.

Wu, G., Siegler, S., Allard, P., Kirtley, C., Leardini, A., Rosenbaum, D., Whittle, M., D’Lima, D.D., Cristofolini, L., Witte, H., Schmid, O. and Stokes, I. (2002). ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion—part I: ankle, hip, and spine. Journal of biomechanics, 35(4), 543-548.

Wu, G., Van Der Helm, F. C., Veeger, H. E. J., Makhsous, M., Van Roy, P., Anglin, C., Nagels, J., Karduna, A.R., McQuade, K., Wang, X., Werner, F.W., and Buchholz, B. (2005). ISB recommendation on definitions of joint coordinate systems of various joints for the reporting of human joint motion—Part II: shoulder, elbow, wrist and hand. Journal of biomechanics, 38(5), 981-992.

Yen, T. Y., and Radwin, R. G. (2007). Usability testing by multimedia video task analysis. Medical Instrumentation: Accessibility and Usability Considerations, 159-172.

Yoo, I. G., Jung, M. Y., Jeon, H. S., and Lee, J. (2010). Effects of Wrist-extension Orthosis on Shoulder and Scapular Muscle Activities during Simulated Assembly Tasks. Industrial health, 48(1), 108-114.

Yun, X., and Bachmann, E. R. (2006). Design, implementation, and experimental results of a quaternion-based Kalman filter for human body motion tracking. Robotics, IEEE Transactions on, 22(6), 1216-1227.

Zhou, H., and Hu, H. (2007). Upper limb motion estimation from inertial measurements. International Journal of Information Technology, 13(1), 1-14.

Zhou, H., and Hu, H. (2010). Reducing drifts in the inertial measurements of wrist and elbow positions. Instrumentation and Measurement, IEEE Transactions on, 59(3), 575-585.

Zhou, H., Hu, H., and Tao, Y. (2006). Inertial measurements of upper limb motion. Medical and Biological Engineering and Computing, 44(6), 479-487.

Zhou, H., Stone, T., Hu, H., and Harris, N. (2008). Use of multiple wearable inertial sensors in upper limb motion tracking. Medical engineering and physics, 30(1), 123-133.

Zhu, R., and Zhou, Z. (2004). A real-time articulated human motion tracking using tri-axis inertial/magnetic sensors package. Neural Systems and Rehabilitation Engineering, IEEE Transactions on, 12(2), 295-302.

Zuckerman, J. D., Kummer, F. J., Cuomo, F., Simon, J., Rosenblum, S., and Katz, N. (1992). The influence of coracoacromial arch anatomy on rotator cuff tears. Journal of Shoulder and Elbow Surgery, 1(1), 4-14.