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Two-Dimensional Measurements of Dynamic Cumulative Spine Loading in Real-Time

By:

Michael James Agnew

A Thesis
Submitted to the Faculty of Graduate Studies and Research through the Faculty of Human Kinetics
in Partial Fulfillment of the Requirements for the Degree of Master of Human Kinetics at the
University of Windsor

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2003

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Abstract

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Michael J. Agnew
University of Windsor, 2003

The purpose of this study was to develop a real time method for documenting the cumulative low back loads present in lifting tasks using a simplified regression based biomechanical model. Following model development, custom software was created and used with an electromagnetic tracking device (Fastrak™) to calculate cumulative loads in real-time. Estimates obtained from the real-time method (cumulative joint compression, anterior joint shear, posterior joint shear) were compared to estimates from a dynamic link segment biomechanical model (GOBER). Estimates obtained by the real-time method were statistically different than the dynamic biomechanical model in estimates of cumulative compression and cumulative anterior shear force ($p = 0.018$, and $p = 0.014$), while measures of cumulative posterior joint shear were shown to be statistically similar ($p = 0.763$). However, further analyses of the real-time method results indicated relative errors of -3.4% ± 3.57%, 4.88% ± 6.47%, and -10.34% ± 33.27% for measurements of cumulative joint compression, anterior joint shear, and posterior joint shear, with cumulative RMS errors of 0.636 kN·s, 34.92 N·s, and 53.73 N·s, respectively. These results indicated that the developed model performs comparably with alternate documented methods used for cumulative load estimation. Therefore it was concluded that the proposed real-time method is valid and could be used for future examinations of cumulative low back loads incurred during sagittal lifting and lowering tasks.
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Chapter 1

INTRODUCTION

Low back problems and work related absence have been reported as an expensive problem in western countries (Delleman, Drost, & Huson, 1992). In fact, low back pain disability is the single most expensive injury that originates in the occupational workplace (Leamon, 1994). It is estimated that in the United States, the annual medical costs for the managing of low back pain is 24 billion dollars and that nearly one quarter of the working population experiences symptoms of low back pain at any given time (Frymoyer & Cats-Baril, 1991). Furthermore, epidemiological evidence has shown that relationships between mechanical loads in a job setting and incidence rates of low back pain do exist in occupational tasks (Chaffin & Park, 1973).

In an attempt to reduce the incidence of occupation related low back pain and/or injury, biomechanical, physiological, and psychophysical data have been used to: 1) Identify the anatomical site(s) of mechanical failure in the lumbar spine, and 2) Compare load criteria observed across case-control groups in a variety of working situations in order to develop baselines, or threshold limit values for given occupational tasks (ex. NIOSH lifting equation) (Delleman et al., 1992).

In the investigation of load limits, peak force magnitudes, mainly related to the compressive force have been used as the only load criteria (Delleman et al., 1992). However, there is a significant amount of research that indicates “shear” forces can contribute greatly towards the development of mechanical failure in the lumbar spine and
therefore, potentially cause the onset of pain in the low back (Yingling & McGill, 1999). These potential dangers of shear forces have also been documented to increase as a result of repeated exertions, thus proving the necessity for this force component to be included in the analysis of trunk kinetics over the course of a work shift, or day (Marras & Granata, 1997).

Although various guidelines have been published in the literature regarding the dangers of peak, or extreme compression and shear forces experienced in the low back, no guidelines exist regarding the cumulative exposure risk of loading in the lumbar spine. The potential dangers of cumulative loading on the tissues of the low back have been identified through both epidemiological (Kumar, 1990; Norman et al., 1998) and in-vitro spine research (Callaghan & McGill, 2001). Commonly, estimation, or measurement of this risk factor is obtained through rectangular integration of the spinal loads (ex. moment, compression, and shear forces at L4/L5) incurred during the completion of a task (Agnew, Andrews, Callaghan, & Potvin, 2002; Andrews & Callaghan, 2003; Callaghan, Salewystch, & Andrews, 2001; Lauder, 2002; Kumar, 1990). By multiplying task cumulative loads by the frequency and duration that the task is performed, the product yields the shift or daily cumulative dose, or exposure level that a given subject, or employee is subjected to over the course of a shift (Norman et al., 1998).

In order to develop threshold limits that are applicable to an entire population, future research in this area requires the collection of data from not only large subject pools, but from a variety of tasks and activities as well. However, the use of traditional biomechanical analyses (ex. video-based biomechanical analysis) to predict and document cumulative low back loads is time consuming and costly to the researcher,
mainly due to the amount of time required to acquire and process data. As a result, a portion of the current research related to cumulative loading has been dedicated towards developing alternate, more efficient methods of quantifying cumulative exposure in the low back. To date, various alternate, more efficient methods have been developed and investigated as potential means of estimating cumulative spine loading, including posture-related questionnaires (Kumar, 1990), reduced sampling rate video-based biomechanical modeling (Andrews et al., 2003; Callaghan et al., 2001; Jager et al., 2000) and via compression-normalized electromyography (Lauder, 2002; Meintjes et al., 1999; Potvin et al., 1990).

An electromagnetic tracking device, such as a Fastrak™ system, enables the collection of positional data from four sensors instantaneously, or in real-time, at a sample rate of 30 Hz. Recent research by Agnew et al. (2002) has indicated that cumulative L4/L5 compression can be quantified accurately through the use of a 4 sensor electromagnetic tracking device and a regression-based static biomechanical model developed by Potvin (1997). Static biomechanical models have been used extensively in the investigation of cumulative loading (Agnew et al., 2002; Andrews et al., 2003; Callaghan et al., 2001; Daynard et al., 2001; Jager et al., 2000; Kumar, 1990; Lauder, 2002; Norman et al., 1998). However, Keown et al. (2001) have shown that the use of static biomechanical modeling underestimates the actual amount of cumulative loading occurring in the low back due to the negation of the inertial forces that occur during the execution of occupational tasks and activities of daily living. Therefore, while alternate efficient means of measuring cumulative loading must be investigated, the optimal method would be one that retained the magnitudes of cumulative load derived from a
dynamic modeling approach, or at least, a quasi-static modeling approach, which has shown to have ~5% relative error compared to a full dynamic model (Keown et al., 2001).

1.1 Statement of the Purpose

The purpose of this study was to develop a lab-based methodology that could quantify dynamic 2-D cumulative spine loading in real-time. In order to achieve this goal, two research objectives were identified and investigated in separate studies. The first objective (Study 1) was to develop a simplified collection method that used regression-based biomechanical models to calculate dynamic forces and moments occurring at the L4/L5 intervertebral joint throughout lifting and lowering tasks. The simplified method required the use of multiple regression models to estimate the inertial weight of the upper body mass, as well as predict the moment arms for both the upper body mass and the hand-held load throughout the completion of the designed tasks. The second objective (Study 2) of this investigation was to evaluate the performance of these models, when driven by data obtained through both video-based and electromagnetically-based instrumentation, as a means of producing accurate estimates of cumulative joint compression and joint shear (both anterior and posterior) forces acting on L4/L5.
1.2 Statement of the Hypotheses

Study 1

1. The proposed regression model will yield statistically similar estimates of the true inertial weight of the upper body mass calculated throughout a given lifting/lowering task.

   A variety of kinematic variables will be measured and used in the creation of the regression model. The equation development component of the study is designed to adequately manipulate the various lifting kinematics that could be adopted in the execution of lifting and lowering tasks. Therefore, the developed model will be robust, and subsequently explain a significant amount of the variance that is attributable to dynamic lifting and lowering.

Study 2

1. The simplified regression model developed in Study 1 will yield statistically similar, cumulative L4/L5 joint compression and anterior/posterior joint shear forces to those obtained from a dynamic rigid link biomechanical model.

   If the estimated inertial weight of the upper body made by the regression model is shown to be statistically accurate to the true measure, it is assumed that the use of this inertial weight model, in concert with regression models used to predict the moment arms for both the upper body and the hand-held load, will yield cumulative spine loads that are statistically similar to those from a dynamic biomechanical model.

2. Real-time measures of cumulative spine loading created with an electromagnetic tracking device and a simplified regression-based model will be statistically similar to those produced by a dynamic rigid link biomechanical model.
Assuming hypothesis 1 (above) is accepted, and that kinematic data obtained through an electromagnetic tracking system are similar to those obtained through video digitization, real-time estimates of cumulative load made by way of the tracking system should be statistically similar to those from a dynamic rigid link biomechanical model.

3. *In the event of significant differences between methods, the differences between both the real-time and video approaches will be due to interaction effects attributable to either load or lift speed magnitude, or a combination of both.*

It is possible that the predictive accuracy of the developed approaches will vary over the different loads and speeds that are to be tested. Therefore, interactions between predictive accuracy and load and speed magnitude could be present within the developed model and thus, further explain measurement error between the criterion measure, and the two methods in which the developed model are to be tested.
Chapter 2

REVIEW OF LITERATURE

2.1 The Lumbar Spine

The following section is a brief review of the following anatomical structures of the lumbar spine: the vertebrae, the intervertebral discs, and the muscles responsible for major movement of the trunk, like a lifting task. For each structure, a brief review of the composition and role each structure plays will be given.

The lumbar region of the spine consists of 5 lumbar vertebrae. Due to the weight of the body mass superior to them, the lumbar vertebrae are the largest and strongest vertebrae in the spine. Each individual vertebra is composed of a body to support the load of the upper body mass, a neural arch that protects the spinal cord, and numerous bony processes that act as attachment sites for muscles, and as articulating surfaces for movement of the vertebrae. Motion of the vertebral bodies is governed by the intervertebral joints (two adjacent vertebrae and an intervertebral disc), which resist compressive force, and the facet joints, which are structured to resist torsion and shearing forces (Bogduk & Twomey, 1987).

The main functions of the intervertebral discs are to allow movement of the spine, dissipate shock transfer between adjacent vertebrae, and aid in the resistance of loads placed on the spine such as compression, shear, and torsion. Intervertebral discs can be divided into two subcomponents, the annulus fibrosus, and the nucleus pulposus.
The annulus fibrosus is composed of a series of 10 to 12 lamellar rings that act as a tensile shell for the disc and blend with the layers of cartilage found on the endplates of superior and inferior vertebrae (Bogduk et al., 1987). Composed of collagen, the fibers of the annulus crisscross in orientation from ring to ring at angles of approximately 30 degrees and are mainly sensitive to rotational strain (Whiting & Zernicke, 1998). The inner component of the disc, the nucleus pulposus, is mechanically designed to resist compression. The nucleus is composed of about 90% water, with the remainder of the material being collagen and proteoglycans. The proteoglycans help cause diffusion of water into the nucleus when it is not loaded, however, throughout the course of time, continual loading of the discs causes the water to be extruded. This extrusion of water can cause discs to shrink in height by as much as 10%, thus compromising their capability to resist and dissipate force (Bogduk et al., 1987).

During lifting, the weight of the upper body mass, and any held load, causes a moment about the axes of rotation of the lumbar vertebrae. Contraction of the lumbar musculature counteracts the moment placed on the lumbar spine and facilitates movement, such as lifting to occur. For gross movements such as this, the major contributing muscles required to extend the lumbar spine are the iliocostalis lumborum and longissimus thoracis, referred to collectively as the lumbar erector spinae. The lumbar erector spinae group of muscles have a line of action that acts obliquely to the lumbar spine. Thus, while a majority of the force generated by the lumbar erector spinae results in compressive force acting on the lumbar vertebrae and discs, shear forces are also produced and act on these structures as well (Bogduk et al., 1987).
2.2 Risks of Cumulative Loading – Epidemiological Evidence

Kumar (1990) first documented the associated risks of cumulative loading through an epidemiological study of institutional aides and their prevalence of low back pain. From a sample of 161 subjects, Kumar (1990) employed a method of posture-based questionnaires to measure the biomechanical forces of compression and shear that were attributable to the daily activities of the recruited subjects. Cumulative loading was estimated through force/time integration and extrapolated into daily, weekly, monthly, yearly, and lifetime estimates of cumulative exposure for both the thoracolumbar and lumbosacral discs. When divided into case/control (pain/no-pain) subject pools, Kumar (1990) documented that in measures of cumulative exposure, institutional aides who reported low pain were exposed, on average, to a significantly higher level of cumulative biomechanical stress than subjects who did not report the occurrence of low back pain.

Similar to Kumar (1990), Norman et al. (1998) identified the associated risks of cumulative loading in an epidemiological study of low back pain reporting. This study attempted to identify the magnitude of peak spine loads, hand loads, trunk kinematics, and cumulative spine loads as predictors of low back pain. Investigated in the automotive industry, this study used 130 randomized controls and 104 cases to identify the increased risk of low back pain as a function of exposure magnitude to the aforementioned risk variable (Norman et al., 1998). In the analysis of work posture, Norman et al. (1998) gathered postural, kinematic, and kinetic data for entry into a biomechanical model by way of video recording and observing the subject for an average of half a shift. Using exploratory factor analysis, Norman et al. (1998) identified that cumulative measures of compression, shear, and moment were independent of all other
recorded variables in terms of explaining a portion of the incidence of low back pain observed in this study, thus proving the potential injury risks and/or onset of low back pain that can be attributable to cumulative loading of the spine.

2.3 Risks of Cumulative Loading – Mechanical Evidence

Repeated application of a load to biological tissues is believed to result in accumulated micro trauma, thereby reducing the tissues’ ability to withstand subsequent loads (Whiting et al., 1998). As a result, failure occurs at much lower loads when loaded repeatedly (Adams & Dolan, 1995; Brinckmann et al., 1988). The strength of a tissue when loaded cyclically is thus a function of both the number of loading cycles (Brinckmann et al., 1988), the magnitude of the load (Brinckmann et al., 1988; Hansson et al., 1987), and the rate at which the tissue is loaded (Yingling et al., 1999). These findings indicate that mechanically, cumulative loading can cause injury in the low back that is independent of the peak loads applied to the system. The following is a review of the peak failure limits of the lumbar vertebrae and the intervertebral discs and the documented cumulative loading that has also been shown to cause tissue failure.

The failure of the lumbar spinal motion segments has been shown to occur at a variety of peak compressive loads. Moore and Garg (1992) have reported failure to occur between compressive loads of 3 to 12 kN. In comparison, Brinckmann et al. (1988) found the ultimate compressive strength of vertebrae to average between 2.1 and 9.6 kN. This damage has been shown to occur at much lower loads when the spine has been loaded repetitively (Adams & Dolan, 1995; Brinckmann et al., 1988; Callaghan & McGill, 2001). When repetitively loaded, Brinckmann et al. (1988) reported a 30%
decrease in compressive strength of the vertebral body. Furthermore, Adams and Dolan (1995) cyclically loaded lumbar motion segments with a compressive force oscillating between 500 and 4000 N in order to mimic the physical demands of a typical work shift. Of thirteen healthy motion segments, five failed at an average peak load of 2240 N, which is far below the peak failure limits reported in the literature. When second sets of specimens were tested, all 29 failed at an average peak load of 3800 N (Adams & Dolan, 1995). Similarly, Callaghan and McGill (2001) have found the initiation of disc herniation to occur with sub-maximal loads as low as 1000 N during high frequency cyclic loading. Although there is evidence that the cumulative effects of force and time can cause injury at sub-maximal load levels, the relationship between the two variables is not known. Through an adaptation of data reported by several researchers, Callaghan (2002) attempted to document the relationship between force and time in the reported failure of in-vitro spine specimens that were cyclically loaded. Assuming an equal relationship between the variables of force and time, a single cumulative threshold value could not be identified. Therefore, it is suggested that force and exposure time are not equally weighted variables and that force plays a dominant role in the development of low back injury and the potential development of pain. This is in agreement with Jager et al. (2000) who propose that a squared, or tetra-powered weighting of force should be incorporated in measurement of cumulative load exposure in the low back. However, since the exact relationship between force and time has not been identified, for the purposes of this study, measures of cumulative loading will involve an equal weighting between these two factors.
2.4 Methodological Issues in Cumulative Load Research

In order to document the amount of cumulative loading that is present in the execution of occupational and non-occupational tasks, an accurate means of measuring this risk factor is required to facilitate the development of threshold limits of cumulative exposure. However, the documentation of cumulative loading in the low back is a research-intensive and laborious undertaking. Therefore, it is necessary that future research of cumulative spine loading is dedicated toward the development of more efficient, yet accurate means of recording data. To date, a variety of measurement methods have been presented in the published literature. While the methodologies employed have shared the common goal of reducing the amount of data to be analyzed; some methods have traded efficiency for accuracy in the documenting of cumulative spine loading.

A recent publication by Callaghan et al. (2001) compared the methodologies employed by Kumar (1990), and Norman et al. (1998) to the assumed criterion measurement of cumulative spine loading via a rigid-link biomechanical model and 30 Hz integration of the force-time history. Kumar (1990) derived cumulative spine loading through the use of posture-based questionnaires that parsed each lifting task into 200 msec intervals (5 Hz). In contrast, Norman et al. (1998) calculated cumulative exposure by finding the peak force value present in a lifting task, and multiplying this value by the total time required to complete the task. The results of the study by Callaghan et al. (2001) indicated that the 5 Hz posture sampling approach employed by Kumar (1990) generated accurate results when compared to the assumed criterion measure (< 5% relative error), while the method used by Norman et al. (1998) resulted in measures of
cumulative loading that contained 70% relative error when compared to the defined criterion approach. Recent work by Andrews et al. (2003) has indicated that posture-sampling rates as low as 2-3 Hz can be used to accurately predict cumulative loading, which in turn reduces the processing demands required to measure cumulative exposure.

Surface electromyography (EMG) has also been assumed to be an efficient means of measuring compression forces throughout the course of a given period of time. Using compression normalized EMG (CNEMG), Potvin et al. (1990), Mientjes et al. (1999) and Lauder (2002), have documented that a large amount of relative error exists with this method when compared to a biomechanical model (~30%). However, when used to develop amplitude probability distribution functions (APDFs), this method generates APDFs that are similar in measure (6.5% relative error) to those derived from a biomechanical model.

Although posture sampling at low frequencies (2-5 Hz) has been proven to be an effective way of reducing the amount of data required to document cumulative loading in the low back, a large amount of processing time is still required to digitize and input kinematic data into a biomechanical model. Research by Agnew et al. (2002) has employed the use of an electromagnetic tracking device to replace video digitization as a means of predicting cumulative spine loading. An electromagnetic tracking device has several advantages over the aforementioned video and posture sampling approaches, as digitized 3-D data can be obtained instantly in real-time. However, due to the sensitivity of the device to metal and electromagnetic distortion, use of this device is limited to a controlled environment, like a laboratory. When used in combination with a regression-based static biomechanical model (Potvin, 1997), Agnew et al. (2002) reported measures
of cumulative compression that were within 3% error of measurements made by a static, rigid link biomechanical model with posture sampling at 30 Hz.

2.5 Static vs. Dynamic Modeling

Static biomechanical modeling accounts for the gravitational forces acting on the limbs modeled, whereas a dynamic approach also incorporates the inertial forces generated by the limbs of the body and any external loads, during the execution of movements, such as a lifting task. By ignoring the inertial properties of the body during a lifting task, static biomechanical models have been shown to underestimate the peak forces occurring in the low back by an average of 19% (McGill & Norman, 1985) and by as much as 60% (Leskinen, et al., 1983). In fact, in the ergonomic evaluation of given lifting tasks, it has been shown that by using a static model, the peak recorded forces that appear to be below ergonomic guidelines, such as the NIOSH MPL (NIOSH, 1981), were actually greater than the safety limit when re-evaluated with a dynamic model (Garg et al., 1982; McGill et al., 1985). However, it must be noted that limits such as the NIOSH MPL were developed based on vertebral failure tests in which the vertebrae, or functional spinal units were loaded as statically as possible, and that similar to other biological tissues, the tissues of the lumbar spine are rate dependent in terms of ultimate strength, such that the maximum tolerances of the tissues are higher when the tissue is loaded dynamically (Yingling et al., 1999). Thus, the applicability of tolerance limits derived via quasi-static failure tests are questionable for comparison of maximum forces that are calculated dynamically via a biomechanical model, as they ignore the rate dependent response of biological tissues.
The underestimation of measurements via static biomechanical modeling compared to dynamic modeling have also been shown to be an issue in the prediction of cumulative spine loading. When estimating the compression\texttimes time integral for controlled lifting tasks, Leskinen et al. (1983) found that the integrals derived from a dynamic model were 28-36\% larger than those derived from a static model. Research by Keown et al. (2001) has also documented that in the estimation of cumulative loading, the use of a static model underestimates the amount of cumulative load derived from a dynamic biomechanical model, and that in the least case, a quasi-static modeling approach should be used, which has shown to be within \(~5\%\) relative error of a full dynamic analysis. Since the mechanical properties of the tissues of the spine have been shown to be rate dependent, future research of cumulative loading, with the ultimate goal of developing threshold limit values of exposure, should incorporate the use of dynamic biomechanical modeling, or at least, a quasi-static approach, in the measurement of cumulative loading in the low back. By doing so, the results of this type of research in the future will document the peak and cumulative magnitudes of loads occurring in the lumbar spine more accurately, while at the same time accounting for the rate at which they were applied.
Chapter 3

METHODOLOGY – STUDY 1

Study 1

As mentioned previously, the purpose of Study 1 was to develop three regression-based biomechanical models that could predict the instantaneous inertial weight of the upper body mass, and the respective moment arms for the upper body mass and hand held load over the course of various lifting/lowering tasks. Prior to the initiation of Study 1 and 2, the methods were reviewed and approved by The Research Ethics Board of the University of Windsor (Appendix A).

3.1 Subjects

Eight male and 8 female subjects (mean height 1.74 m ± 0.08 m, mean mass 74.94 kg ± 11.74 kg,) from a university population volunteered to participate in this component of the study. All subjects were healthy with no prior history of low back pain and were required to read an information form and complete a consent form prior to participation (Appendix A).
3.2 Experimental Apparatus

The following sections identify and explain the instrumentation and apparatus used for both Study 1 and Study 2. Additional instrumentation, used exclusively in Study 2, is defined and explained later in the document within the methodology section for Study 2.

3.2.1 Lifting Platform

Due to the use of an electromagnetic tracking device later in Study 2, a wooden lifting platform was constructed for use in both studies (Figure 1). The main function of the lifting platform was to increase the distance between the Fastrak™ hardware and the steel-reinforced floor that can increase the error in the positional measurements made by the electromagnetic tracking device. The platform was constructed of spruce plywood and lumber, measuring 200* 70 * 50 cm. A 140 cm shelf was also constructed out of spruce lumber and plywood for use in this study. Electrical trigger switches were constructed and placed on the floor of the platform and on top of the shelf for use in Study 2. Further detail regarding the design and use of the switches is described in the methodology of Study 2. It was assumed that the design of the platform and its components did not alter the normal lifting mechanics displayed by a subject lifting from true floor level.
Figure 1. The lifting platform used for both studies. The black cloth visible in this picture was used to help in the reflection and contrast of reflective markers required for both Study 1 and Study 2. Also visible within this photograph is the mounting location of the Fastrak™ source cube and one of the trigger switches used in Study 2.

3.2.2 Video Motion Analysis

Both studies required the collection of two-dimensional kinematic data. The collection of these data was facilitated through the use of video motion analysis software (APAS, Ariel Dynamics, San Diego, California). The video data collection protocol required the use of reflective markers that were used in the auto-digitization of the individual joint coordinates by way of video motion analysis software. The markers used were half spheres made of Styrofoam™ that were covered in reflective tape (Figure 2). The markers measured 2.54 cm in diameter by 1.25 cm in height and were illuminated by way of a spotlight placed perpendicular to the reflective surfaces of the markers. A digital video camera (Canon Optura 200MC, Canon Incorporated, Japan) was positioned perpendicular to the lifting platform on the right hand side at a distance of 5 m. The
video camera recorded subject's performances to Mini-DV videotape at a sample rate of 30 frames per second.

Prior to collection the 2-D camera field of view was scaled and calibrated using a calibration jig with 4 markers placed at a known distance (Figure 3). All lifts were performed in the sagittal plane and were assumed to be symmetrical. Reflective markers were attached to the skin at the following joints of each subject on the right hand side of the body; the distal head of the 3rd metacarpal (hand), wrist, elbow, shoulder, ear canal, ankle, as well as fins that were dorsally fixed at the C7/T1 and L4/L5 intervertebral joints (Figure 5). The fins were constructed of a rigid material and were used to linearly translate the external marker co-ordinates into internal 2-D co-ordinates of the C7/T1, and L4/L5 intervertebral joints. Two reflective markers were attached to each fin 6 cm apart; the proximal marker was located 5.5 cm away from the skin. The vector drawn between the two markers on each fin was translated into the body a known distance to predict the internal 2-D co-ordinates for the axes of rotation of the two intervertebral joints. The distances used for translation were 50% of the total depth of the trunk at C7/T1, and 43% of the total trunk depth at L4/L5 (McGill et al., 1988). The researcher made the depth measurements with each subject lying in a supine position prior to the beginning of the collection session.
Figures 2 & 3. The reflective markers and fins (2) and calibration jig (3) used in both studies. The calibration square had 4 markers spaced 85 cm apart in order to calibrate the camera field of view.

Figure 4. The joint marker placements used for video digitization.

As mentioned earlier, the reflective joint markers captured by video were automatically digitized in 2-D using APAS software. APAS Video-capture hardware and software were
also used to convert the video data into digital form on a Pentium III PC compatible computer. Following the digitization process, data were digitally low pass filtered at a cut-off frequency of 6 Hz in order to eliminate any signal noise or artefact that was created throughout the digitizing process. Digitized data from all trials were individually written out in ASCII format to computer text files that were used by statistical and biomechanical software applications later in this study.

3.3 Data Collection

Subjects were required to perform lifting and lowering tasks from the platform floor to the shelf that was constructed. Trials were varied by load (3.5, 8, 12.5 kg), lift speed (slow, normal, and fast), and lift style (squat or stoop) for a total of 18 lifts and 18 lowers per subject. For the slow and fast lift speed conditions, subjects were told to perform the tasks as fast or as slow as possible without placing themselves at risk of injury. The purpose of these manipulations was to elicit the greatest number of variations possible for any of the dynamic variables that were analyzed during the completion of a lifting and lowering task. By doing so, it was assumed that the developed equations would be robust, based on the data from which they were created, and therefore govern the inter- and intra-subject variability typical within any population of subjects when used. As a result, for the three lift speed conditions (slow, normal, fast), average lifting speeds of 0.358 m/s ± 0.07 m/s, 0.452 m/s ± 0.06 m/s and 0.588 m/s ± 0.08 m/s were observed, respectively. The conditions were presented in randomized blocks according to load to prevent order effects that could have confounded the study. While subjects did lift and lower loads of varying mass throughout all of the conditions, all collected data were
processed as if subjects did not have any mass in their hands at all. By assuming zero force in the hands, the dynamic and static contributions of the upper body could be calculated for each recorded task. Therefore, for clarification, hand loads were used in this portion of the study to illicit lifting characteristics and dynamics that were true to the tasks observed. It was assumed that these characteristics would not have been observed if subjects had performed the conditions without any load in the hands.

Subjects were given an auditory cue to begin each condition, upon which a trigger light was initiated. The trigger light was visible within the field of view of the video camera and used during the video editing process to mark the beginning and end of a trial. Data obtained from the collection session were used towards the development of three regression-based equations that predict the moment arm of the upper body centre of mass, the moment arm of the load, and the instantaneous inertial weight of the upper body mass.

3.4 Data Analysis

The developed models were based on research by Potvin (1997), who developed regression equations to predict the moment arms of the upper body centre of mass and load for any given posture using NIOSH (NIOSH, 1981) equation measures of H (horizontal displacement of the load in the hand to the ankle (m)), V (vertical displacement of the load with respect to the floor (m)), as well as a measurement of the trunk angle TA with respect to the horizontal (rads) (Figure 5). Because the moment arm equations published by Potvin (1997) are relative to the L5/S1 joint, data collected in Study 1 were used to develop moment arms equations relative to L4/L5 for the upper
body and load, respectively, since the biomechanical model being used as the criterion throughout this study measured forces and moments relative to L4/L5.

![Diagram](image.png)

**Figure 5.** Schematic representation of the regression model approach that was used in this study (from Potvin (1997)). $MA_{CM}$ represents the moment arm of the upper body mass (m), $W_{ICM}$ is the weight of the upper body (N), $MA_{LD}$ is the moment arm of the load (m), $W_{ILD}$ is the weight of the load (N).

3.4.1 Development of Moment Arm Centre of Mass Equation

Co-ordinate data collected from the testing sessions were input into a static biomechanical model (GOBER, University of Waterloo, Waterloo, Ontario), from which, a L4/L5 moment/time history could be derived for each recorded lifting and lowering task. The L4/L5 moment/histories obtained for each subject were divided by the subject’s upper body weight (53.6%, 50.4% total body mass, males and females, respectively, where body weight (N) = body mass (kg) * 9.81 m/s^2, from Potvin (1997)) to yield horizontal moment arm/time histories of the upper body centre of mass for each condition. Similarly, for each task, co-ordinate data were also used to calculate time history data for measures of H, V, TA, and cos (TA). As mentioned previously, video
data were collected at 30 Hz, or in other words, 30 static postures per second. Therefore, the time series data collected from all subjects represented the total amount of static postures to be used in the regression analysis, whereby measures of $H$, $V$, $TA$, and $\cos (TA)$ could be manipulated to accurately predict the dependent variable: the moment arm length for the upper body centre of mass for any given posture.

### 3.4.2 Development of Moment Arm Hand Load Equation

For the development of this equation, the same approach was used as in the development of the upper body centre of mass equation. However, while the same independent variables were used in the analysis ($H$, $V$, $TA$, $\cos (TA)$), co-ordinate data were used to predict the true moment arm of the hand held load by calculating the horizontal displacement of the hand marker ($3^{rd}$ metacarpal) with respect to the estimated internal location of the L4/L5 intervertebral joint for each captured video frame.

### 3.4.3 Development of Equation to Predict Inertial Weight of the Upper Body

The final objective for Study 1 was to develop a regression-based model that predicted the inertial weight of the upper body for any recorded frame of a lifting/lowering task. For each condition, co-ordinate displacement data were input into the GOBER biomechanical software package to yield both a static and dynamic compression force/time history. By dividing the dynamic compression force/time history by the static compression force/time history for each condition, an inertial upper body weight could be calculated for each recorded posture. Theoretically, the inertial weight calculated for each observed posture quantified a total upper body weight that, when used
within a static biomechanical model for each observed posture, would yield the same measures of force at L4/L5 as those calculated using a dynamic biomechanical approach.

Through multiple stepwise regression, an analysis was conducted to determine if a significant portion of the variance observed in the measures of inertial upper body weight could be explained using subjects' static upper body weight, H, V, TA, and the 2nd derivatives of these values (which are equivalent to the horizontal acceleration of the hand with respect to the ankle (H''), the vertical acceleration of the hand with respect to the ground (V''), and the angular acceleration of the trunk (TA'')) as independent variables.
4.1 Regression Equation – Moment Arm Upper Body Centre of Mass

From the collection sessions, criterion and independent variable data were derived for 53 547 postures (video frames) and input into statistical software to conduct a multiple regression analysis on the data obtained. From the regression analysis, the following equation was produced:

\[
MA_{CM} = -0.012 + 0.113*(H) - 0.015*(V) + 0.009*(TA) + 0.295*(\cos(TA)) \quad [1]
\]

Where:

- \(MA_{CM}\) is the moment arm of the upper body centre of mass (m)
- \(H\) is the horizontal displacement between the load (3rd metacarpal marker) and the ankle (m)
- \(V\) is the vertical displacement of the hand held load (m)
- \(TA\) is the angle of the trunk with respect to the horizontal (rads)
- \(\cos(TA)\) is the cosine of the measured trunk angle \(TA\)

The developed model was highly correlated to the criterion measurement for the length of the upper body centre of mass moment arm (Figure 6), with a low root mean square (RMS) error \(R^2 = 0.992\), RMS error = 0.01 m) and a significance less than \(p = 0.0001\).
Figure 6. X/Y scatter of the predicted model (Y) versus the criterion centre of mass moment arm (m) values obtained through the GOBER software (X).

4.2 Regression Equation – Moment Arm Hand Load

Using the same data set applied in Equation 1, a similar approach was used in the development of a regression model to estimate the moment arm of the hand held load for any given posture within a lifting/lowering task. For this analysis, the same independent variables were used as in the development of Equation 1. However, these variables were now used to predict the moment arm of the load in the hands, the criterion of which being the calculated horizontal distance between the hand marker (3rd metacarpal) and the internal location of the L4/L5 joint. From the regression analysis, the following equation was developed:
\[ MA_{LD} = -0.183 + 0.996^*(H) - 0.004^*(V) + 0.08^*(TA) + 0.267^*(\cos(TA)) \]  \[2\]

Where:

- \( MA_{LD} \) is the moment arm of the hand held load (m)
- \( H \) is the horizontal displacement between the load (3\textsuperscript{rd} metacarpal marker) and the ankle (m)
- \( V \) is the vertical displacement of the hand held load (m)
- \( TA \) is the angle of the trunk with respect to the horizontal (rads)
- \( \cos(TA) \) is the cosine of the measured trunk angle TA

Similar to the results of Equation 1, the 2\textsuperscript{nd} developed model, Equation 2, also explained a significant \( p < 0.0001 \) amount of the observed variance in load moment arm length \( R^2 = 0.98, \) RMS error = 0.029 m) and is depicted in Figure 7.
4.3 Regression Equation – Inertial Upper Body Weight

Following inspection and removal of outliers, 51,567 postures were used in the development of the inertial upper body weight regression equation. To clarify, any posture (data frame) which yielded a dynamic measure (H”, V”, or TA”) greater than 3 standard deviations away from the mean of each respective acceleration data set was defined as an outlier (Pedhazur, 1997). Therefore, approximately 3.7% of the recorded data frames were removed from this component of the analysis. In the development of the equation, the accelerations were low pass filtered using a dual pass Butterworth filter. A variety of different filter cut-off frequencies were tested to identify which treatments
produced trends within the acceleration data sets that could explain the most variance of the dependent variable. Following these analyses, the following equation was found to explain the greatest amount of variance in the inertial weight of the upper body mass:

\[
\text{Inertial}_{\text{w,tcm}} = -26.45 + 0.991*(\text{Static}_{\text{w,tcm}}) - 16.52*(V) + 7.15*(V'') + 79.2*(H) + 15.5*(TA) + 17.01*(TA'')
\]  

[3]

Where:
- \(\text{Inertial}_{\text{w,tcm}}\) is the inertial weight of the upper body mass for any given posture (N)
- \(\text{Static}_{\text{w,tcm}}\) is the static weight of the upper body mass - upper body mass \(* 9.81\text{m/s}^2\) (N)
- \(V\) is the vertical displacement of the hand held load (m)
- \(V''\) is the 2\text{nd} derivative of \(V\) (m/s\(^2\))
- \(H\) is the horizontal displacement of the load/hands with respect to the ankle (m)
- \(TA\) is the angle of the trunk with respect to the horizontal (rads)
- \(TA''\) is the 2\text{nd} derivative of \(TA\) (rads/s\(^2\))

The final cut-off frequencies chosen for the acceleration variables were: 2 Hz for \(V''\), and 1.5 Hz for \(TA''\) as they explained the greatest portion of variance attributable to the inertial weight of the upper body. By using these cut-off frequencies within the above equation, a significant \((p < 0.0001)\) amount of the variance \((R^2 = 0.92, \text{RMS error, 28.45 N})\) was explained via the developed model (Figure 8).
Figure 8. X/Y scatter of the predicted inertial weight model (Y) versus the criterion inertial weight (N) measures of the upper body (X) obtained from the GOBER biomechanical model.
Chapter 5

METHODOLOGY – STUDY 2

Study 2

Study 2 was designed to validate the use of the models created in Study 1 for the measurement of the cumulative L4/L5 compression, and anterior/posterior shear forces incurred during lifting and lowering tasks. Using cumulative load measures derived from a dynamic biomechanical model (GOBER, University of Waterloo, Waterloo, Ontario), the accuracy of the developed models was tested using both video-based data and data obtained in real-time by way of a Fastrak™ electromagnetic tracking device.

5.1 Subjects

Eight healthy subjects (4 male, 4 female, mean height 1.78m ± 0.11m, mean mass 75.87 kg, ± 20.14 kg) without history of low back pain agreed to participate in the study. Subjects who participated in Study 1 were not allowed to volunteer for Study 2. All subjects were from a university population and were required to read and sign the same consent form used in Study 1 (Appendix A).
5.2 Additional Experimental Apparatus

All of the apparatus and instrumentation used in Study 1 were also required for Study 2. The following is a description of the additional apparatus and instruments required for the completion of Study 2.

5.2.1 Electromagnetic Tracking Device

Two-dimensional kinematic data were also collected using a Fastrak™ electromagnetic tracking device (Polehemus Technologies, Colchester, Vermont). The system used consisted of an electromagnetic transmitter cube and 4 sensors (Figure 9). The transmitter cube produced an electromagnetic field whereby each of the 4 sensors were located and measured in terms of orthogonal XYZ position relative to the source cube. The manufacturer specifies an accuracy of less than 2% error for any measurement within 75 cm of XYZ position relative to the source cube. Therefore, to maintain measurement accuracy, the transmitter cube of the Fastrak™ was positioned and aligned with the lifting platform in a manner so that the greatest displacement in any plane for a sensor was ≤ 75 cm during the completion of a task. Following the alignment of the transmitter cube, a series of measurements were taken within the volume that the sensors were to be displaced during the study. The results of the measurements were within the manufacturers specifications, therefore it was concluded that the system was aligned correctly to produce the best possible measurement accuracy.
Figure 9. The Fastrak™ device used to capture real-time 2-D coordinate data.

The four Fastrak™ sensors were placed at the distal head of the 3rd metacarpal, the ankle, and dorsally on the skin at the site of the C7/T1 and L4/L5 intervertebral joints. The internal 2-D location of the intervertebral joints was also estimated from the Fastrak™ data. The procedure required estimated trunk depths for both the cervical and lumbar joints, as well as a measurement of the angle, relative to the horizontal, that existed between the two spinal sensors for any given posture.

Communication between the Fastrak™ device and a Pentium IV PC computer was made via a RS-232 communications port built into the Fastrak™, and a standard 9-pin communications port fixed to the computer set to transfer data at a rate of 115 200 baud. All commands and data acquisitions were made via custom Lab VIEW software (National Instruments, Austin, Texas) programmed specifically for this study. All positional measurements made by the Fastrak™ device were low pass filtered using a dual pass Butterworth filter with a final cut-off frequency of 6 Hz. Similar to the video
analysis, the Fastrak™ data were written out to individual ASCII text files to be used for further analysis.

5.2.2 Electrical Trigger Switches

The real-time method developed in Study 2 required the monitoring of the instant when a load had left contact with the ground, as well as the instant the load had been placed on the shelf and was no longer in the hands. This was achieved using electrical trigger switches that were built into the platform floor and on top of the platform shelf. The switches were spring loaded and displaced two metal contacts a distance of 0.5 cm to either complete or break a 5 volt circuit. Voltage data travelling through the circuit were sampled at 30 Hz and converted to digital form using a 12-bit A/D card (National Instruments, Austin, Texas) for use in the Lab VIEW software designed for this study.

5.3 Data Collection

Similar to Study 1, subjects were required to perform lifting and lowering tasks while standing on the lifting platform. Trials were again varied by load (3.5, 8, 12.5 kg), and lift speed (slow, normal, fast) for a total of 9 lifts and 9 lowers per subject (N = 144). For each condition, subjects were instructed to lift with a freestyle approach, as opposed to the forced stoop and squat styles used in Study 1. The purpose of these manipulations was to test and evaluate the performance of the developed models across a range of lifting dynamics, and physical demands. As a result, the lift speeds observed in Study 2 were similar in magnitude to those observed in Study 1 with estimated lifting speeds of 0.345 m/s ± 0.04 m/s, 0.433 m/s ± 0.03 m/s, and 0.504 m/s ± 0.05 m/s respectively for the
slow, normal, and fast conditions. The conditions were presented in randomized blocks according to load to prevent order effects that could have confounded the study.

Subjects were given an auditory cue to begin each condition, upon which a trigger light was initiated to synchronize video and Fastrak™ data acquisition. Through the use of the trigger switches to identify when the load was in/out of the hands, a dynamic force time history could be derived for the load in the hands through the use of the Fastrak™ sensor fixed to hand and the following equation:

\[ \text{Inertial}_{\text{LD}} = m(a + g) \]  

Where:
\( W_{\text{LD}} \) is the inertial weight of the load in the hands (N)
\( m \) is the static load (kg)
\( g \) is the acceleration due to gravity, 9.81 m/s²
\( a \) is the additional acceleration placed on the load, assumed to be the vertical acceleration of the hand, or the 2\(^{\text{nd}}\) derivative of \( V \), \( V'' \), which was dual pass, low pass filtered at a resulting cut-off frequency of 10 Hz

Therefore, for each condition, 2-D co-ordinates were collected from the video-based measurement system, and the Fastrak™ device, as well as the dynamic forces calculated to occur at the hands throughout the completion of the lifting and lowering tasks.

The video-based co-ordinate data and the dynamic hand force/time histories were input into the GOBER biomechanical software package to produce dynamic L4/L5 force/time histories of compression (N), posterior joint shear (N), and anterior joint shear (N) for each task. Similarly, L4/L5 force/time histories were derived from video-based co-ordinate data and Fastrak™ data using the following procedures:
\[ M_{L4/L5} = (Wt_{CM} \times MA_{CM}) + (Wt_{LD} \times MA_{LD}) \]  

Where:  
\( M_{L4/L5} \) is the dynamic external moment placed on L4/L5 for each data frame (positive value indicating a flexor moment, negative value indicating an extensor moment)  
\( Wt_{CM} \) is the inertial weight (N) of the upper body mass calculated using Equation 3  
\( MA_{CM} \) is the predicted moment arm (m) of the upper body center of mass calculated using Equation 1  
\( Wt_{LD} \) is the dynamic, or inertial weight (N) of the load in the hands calculated using Equation 4  
\( MA_{LD} \) is the predicted moment arm (m) of the load in the hands, calculated using Equation 2

Using both measurement devices (video and Fastrak™ co-ordinate data), a dynamic external moment was calculated for each captured frame of data during the completion of the conditions presented within the study. The external dynamic moment calculated from Equation 5 was resolved into internal forces acting on L4/L5 using a single muscle equivalent approach. The single muscle equivalent approach assumed one moment arm to represent the line of action of the muscles posterior or anterior of the joint, in order to counteract the external moment placed on the body for any given posture. From this analysis it was possible to calculate a force/time series for the performance of each presented condition for measures of L4/L5 compression, anterior and posterior joint shear forces using the following equations (adapted from Potvin, 1997):

For External Flexor Moments:

\[ F_c = [M_{L4/L5} + (0.06 + (\cos 5.3^\circ))] + \sin (TA) \times (Wt_{LD} + Wt_{CM}) \]  

Where:  
\( F_c \) is the compressive force (N) acting on L4/L5 for each data frame  
\( M_{L4/L5} \) is the dynamic external moment (Nm) acting on L4/L5, calculated using Equation 5  
\( TA \) is the angle of the trunk with respect to the horizontal (rads)
$W_{tLD}$ is the dynamic force (N) acting on the hands for each data frame, calculated using Equation 4
$W_{tCM}$ is the inertial weight (N) of the upper body mass for each data frame, calculated using Equation 3

Note: The moment arm length used to represent the lumbar musculature posterior to LA/L5 was the same as that used by the GOBER biomechanical software.

$$S_{RXN} = \cos (TA) \times (W_{tLD} + W_{tCM}) \quad \text{[7]}$$

Where:
$S_{RXN}$ is the reaction shear force (N) acting on L4/L5 for each data frame
$TA$ is the angle of the trunk with respect to the horizontal (rads)
$W_{tLD}$ is the dynamic force (N) acting on the hands for each data frame, calculated using Equation 4
$W_{tCM}$ is the inertial weight (N) of the upper body mass for each data frame, calculated using Equation 3.

$$S_{INT} = -\left\{ (M_{L4/L5} \div (0.06 \div (\cos 5.3^\circ))) \times 0.093 \right\} + S_{RXN} \quad \text{[8]}$$

Where:
$S_{INT}$ is the total joint shear force (N) (positive value indicates anterior joint shear, negative indicates posterior joint shear) acting on L4/L5 for each data frame
$M_{L4/L5}$ is the dynamic external moment (Nm) acting on L4/L5, calculated using Equation 5
$S_{RXN}$ is the reaction shear force (N) acting on L4/L5 for each data frame, calculated using Equation 7

Note: The lumbar musculature has been shown to have an oblique angle of pull on the L4/L5 joint. Therefore, the lumbar musculature also causes posterior joint shear force (N) that has been shown to be proportional (9.3%) to the amount of joint compression force (N) required to balance the external moment (Potvin et al., 1991).

For External Extensor Moments:

$$F_c = [M_{L4/L5} \div (-0.045)] + \sin (TA) \times (W_{tLD} + W_{tCM}) \quad \text{[9]}$$

Where:
$F_c$ is the compressive force (N) acting on L4/L5 for each data frame
$M_{L4/L5}$ is the dynamic external moment (Nm) acting on L4/L5, calculated using Equation 5
$TA$ is the angle of the trunk with respect to the horizontal (rads)
$W_{tLD}$ is the dynamic force (N) acting on the hands for each data frame, calculated using Equation 4
$W_{tCM}$ is the inertial weight (N) of the upper body mass for each data frame, calculated using Equation 3

Note: The moment arm length used to represent the lumbar musculature anterior to L4/L5 was the same as that used by the GOBER biomechanical software.
\[ S_{INT} = \cos(TA) \times (W_{LD} + W_{CM}) \]  \[10\]

Where:
- \(S_{INT}\) is the total joint shear force (N) (positive value indicates anterior joint shear, negative indicates posterior joint shear) acting on L4/L5 for each data frame
- \(TA\) is the angle of the trunk with respect to the horizontal (rads)
- \(W_{LD}\) is the dynamic force (N) acting on the hands for each data frame, calculated using Equation 4
- \(W_{CM}\) is the inertial weight (N) of the upper body mass for each data frame, calculated using Equation 3

Note: The moment arm used to represent the musculature anterior to L4/L5 was modeled by the GOBER software to have an angle of pull perpendicular to the L4/L5. This meant no additional shear forces were created as a result of the line of action used to represent the musculature anterior to L4/L5. Therefore, for external extensor moments, the total joint shear forces calculated were equivalent to the reaction shear forces calculated for each data frame.

5.4 Data Analysis

Each equation was programmed into custom Lab VIEW software so that coordinate data obtained via video digitization could be used with the developed regression models. Additional software was created to facilitate the real-time acquisition of coordinate data from Fastrak™ for direct use within the developed models. Therefore, for each recorded trial, dynamic force/time histories of L4/L5 compression (N), posterior joint shear (N), and anterior joint shear (N) were derived using the GOBER biomechanical software package, and the developed regression approach, using data obtained by both video digitization and the Fastrak™ device.

Estimates of cumulative load (compression, posterior joint shear, and anterior joint shear (N.s)) incurred during each task were calculated for the three methods (GOBER, the developed model, driven by video-based co-ordinate data (Video method), and the developed model, driven by Fastrak™ data (Fastrak™ method)) through
rectangular integration of the calculated force/time histories. Using a 3 factor (Method x Lift Speed x Load) repeated measures analysis of variance (ANOVA), the statistical agreement between the three methods was investigated (Figure 11). Significance was set at $\alpha = 0.05$. Post hoc analyses were conducted using paired t-tests that were adjusted using the Bonferroni correction factor to eliminate family-wise error within the comparisons.

To further evaluate the measurement accuracy of the developed model within each lifting condition, the force/time histories derived from both the video and Fastrak™ co-ordinate data were compared against those derived via the GOBER biomechanical model. Assuming the GOBER model to be the criterion measure, instantaneous and cumulative RMS errors of the two "simplified" model approaches were calculated for each trial, and compared when collapsed across conditions (Load, and Lift Speed), and overall. Relative errors between methods were also calculated and are presented in the results.

![Diagram](image)

**Figure 10.** Schematic of the $3 \times 3 \times 3$ repeated measures ANOVA model used for statistical analysis.
Chapter 6

RESULTS – STUDY 2

The data presented in this chapter is broken into three sections dedicated to each of the three dependent measures; cumulative compression, cumulative anterior joint shear, and cumulative posterior joint shear. Within each section, the results of the ANOVA are presented, as well as other comparative procedures, such as relative error calculations and RMS error values (both instantaneous and cumulative) for each of the developed measurement methods with respect to the GOBER model.

6.1 Cumulative Compression

For estimates of cumulative compression, a significant main effect was found for Lift Speed ($p < 0.001$). Post hoc analyses for Lift Speed revealed that each of the three levels was significantly greater than the next, such that cumulative compressions incurred during slow lifting were larger than the magnitudes incurred with the normal speed condition ($p < 0.001$), which in turn were larger than those incurred within the fast lifting conditions ($p < 0.001$) (Figure 11).

A significant Method*Load interaction effect was also identified ($p < 0.001$). This interaction effect indicated that the main effects of Load and Method were not independent of each other, and therefore, the variance within each level of one of the factors was modulated differently at varying levels of the other factor, and vice versa. Since the focus of this study was to compare the statistical similarity of the three levels of
Method (GOBER model, Video method, Fastrak™ method), post hoc analyses of the interaction effect were performed to evaluate the agreement between the three methods at each level of Load. Comparisons within Method for each level of Load were tested at \( p = 0.006 \) (as opposed to \( p = 0.05 \)), due to the Bonferroni correction procedure for multiple comparisons, for which in this case there were nine.

For the 3.5 kg Load condition, cumulative compression magnitudes derived from the GOBER and Video method were statistically similar, whereas estimates derived via the Fastrak™ method were statistically smaller in magnitude. For the 8 kg Load condition, estimates derived from the GOBER model were statistically greater than both the Video method and the Fastrak™ method, which were shown to be statistically equal in magnitude. Finally, for the 12.5 kg condition, estimates derived from the GOBER model were statistically greater than both the Video method and the Fastrak™ method, which were again shown to be statistically similar (Figure 11).
Figure 11. The overall main effect of Lift Speed and the observed Load*Method interaction. Arrows represent significant differences between levels of method across loads ($p < 0.006$) and the overall main effect of Lift Speed ($p < 0.001$).

Overall, in terms of relative error, the Video method and Fastrak™ method consistently underestimated cumulative compression magnitudes derived by the GOBER model, by $3.4\% \pm 3.57\%$, and $1.67\% \pm 5.91\%$, respectively (Figure 12).
**Figure 12.** The observed relative error (%) of the Video method and Fastrak™ method when compared to the GOBER model across all conditions, and overall, for cumulative compression.

With respect to the RMS error, both methods were equally modulated across the Lift Speed factor, which interacted with the Method factor ($p < 0.001$), indicating significant increases of RMS error with both the Video method and Fastrak™ method as lifting speed was increased (Figure 13). Post hoc analysis of the interaction indicated that a significant difference between the two methods existed within the slow lifting speed ($p < 0.001$), whereas no difference existed between models for the remaining levels of the Lift Speed factor. The increase in RMS error by the two developed methods is evident in Figures 14 and 15, which are force/time histories calculated during a normal and fast lift speed condition.
There was no main effect of Load in terms of RMS error. Overall, when collapsed across Load and Lift Speed, the average RMS errors for the Video method and Fastrak method were 167.91 N ± 80.68 N and 154.83 N ± 78.97 N, respectively (Figure 13), with overall cumulative RMS errors of 0.57 kN·s for the video method and 0.636 kN·s for the Fastrak™ method.

**Figure 13.** The compression RMS error (N) attributable to each of the developed methods with respect to the GOBER model. The * indicates a significant difference between the Video method and Fastrak™ method ($p < 0.001$), arrows indicate the main effect of Lift Speed ($p < 0.01$, $p < 0.001$).
Figure 14. Example of a L4/L5 compression force/time history (N) derived from each of the three methods within a normal lifting condition.

Figure 15. Example of a L4/L5 compression force/time history (N) derived from each of the three methods within a fast lifting condition. Note the increase in RMS error for the two developed models when compared to Figure 14.
6.2 Cumulative Anterior Joint Shear

Significant main effects of Method, Load, and Lift Speed were identified from the ANOVA ($p < 0.001$, $p < 0.001$, and $p < 0.05$, respectively), as were significant Method*Lift Speed and Method*Load interactions ($p < 0.001$ for both). Post hoc comparisons of the interaction effects indicated that, for each level of Lift Speed and Load, the three models were significantly different from each other, such that Video method measures were greater than Fastrak measures ($p < 0.001$), which in turn were significantly greater than measures derived from GOBER ($p < 0.001$) (Figure 16). While the three methods were different than each other at each level of Lift Speed and Load, the differences between the two developed methods and the GOBER model increased in magnitude as Lift Speed and Load were increased.

![Graph showing Cumulative Anterior Joint Shear](image)

**Figure 16.** Observed cumulative anterior shear magnitudes (N.s) derived by the three methods across Load and Lift Speed. Significant differences were present between the three methods across both factors. Differences increased with both Load and Lift Speed (all $p < 0.001$).
In terms of relative error, the Video method and Fastrak™ method overestimated measures of cumulative anterior shear derived by the GOBER model, by an average of 10.05% ± 7.29%, and 4.88% ± 6.47%, respectively (Figure 17).

**Figure 17.** The observed relative error (%) of the Video method and Fastrak™ method when compared to the GOBER model across all conditions, and overall, for cumulative anterior shear.
A comparison of the RMS error existing within and between the Video method and Fastrak™ method indicated significant main effects of Method and Load (\(p < 0.01\) and \(p < 0.001\), respectively). The presence of no interaction effects indicated that the levels of Load had a similar effect on both methods, causing an increase in RMS error with an increase in the load lifted. The RMS error associated with the Video method and Fastrak™ method remained statistically different across all conditions, with overall RMS errors of 25.58 N ± 9.73 N and 23.63 N ± 8.48 N, respectively (Figure 18). In terms of cumulative RMS error, overall cumulative RMS errors of 32.31 N·s and 34.92 N·s were found for the Video and Fastrak™ methods, respectively.

![Figure 18](image.png)

**Figure 18.** Observed RMS error (N) for the developed methods when tested across Load, Load Speed, and overall for estimates of cumulative anterior joint shear. Arrows indicate the main effect of Load on RMS error within methods (\(p < 0.001\) for both). Across each factor, and overall, RMS errors between the two methods were significantly different (\(p < 0.01\)).
6.3 Cumulative Posterior Joint Shear

For cumulative posterior joint shear, a significant main effect was found for Load, such that observed magnitudes of cumulative posterior joint shear significantly decreased as load was increased ($p < 0.001$). A significant Lift Speed*Method interaction ($p < 0.001$) indicated that differences between methods varied as a function of Lift Speed (Figure 19). Post hoc analyses of the interaction revealed that for slow lifting speed, measures derived from the Video method were statistically similar to the Fastrak™ method ($p = 0.380$), which in turn was statistically similar to the GOBER model ($p = 0.295$). However, within this level of Lift Speed, the Video method significantly underestimated the GOBER model ($p < 0.001$). For normal lifting speeds, the same relationship was found, such that the Video method significantly underestimated the amount of cumulative posterior joint shear derived by the GOBER model ($p < 0.001$), while at the same time, being similar to the Fastrak™ method ($p = 0.560$). In addition, the Fastrak™ method was also found to be statistically similar to the GOBER model for normal lifting speeds in terms of cumulative posterior joint shear ($p = 0.058$). For fast lifting, the Video method and Fastrak™ method were statistically similar ($p = 0.181$), however, both methods underestimated the cumulative loads derived by the GOBER model within this condition (both $p < 0.001$) (Figure 19).

In terms of relative error, the Video method and Fastrak™ method were both shown to underestimate the GOBER model, with overall relative errors of -15.11% ± 22.20% and -10.34% ± 33.27%, respectively (Figure 20). For RMS error, both the Load*Method, and Lift Speed*Method interactions were significant ($p < 0.05$ for both).
Post hoc analyses revealed a significant difference in RMS error between the Video method and the Fastrak™ method for the 8 kg Load condition, and the slow Lift Speed condition (both $p < 0.001$) (Figure 21). Overall, the Video method and Fastrak™ method had RMS errors of $9.39 \text{ N} \pm 3.98$, and $9.39 \text{ N} \pm 3.72 \text{ N}$, respectively, with overall cumulative RMS errors of 58.62 N·s and 53.73 N·s for each method.

**Figure 19.** Model estimates of cumulative posterior joint shear (N.s) across levels of Load and Lift Speed. Arrows indicate significant differences between methods and across Loads ($p < 0.001$).
Figure 20. The relative error (%) existing between methods across Load, Lift Speed, and overall in estimates of cumulative posterior joint shear.

Figure 21. The RMS errors (N) existing between methods across Load, Lift Speed, and overall in estimates of cumulative posterior joint shear. * indicates significant difference at $p < 0.001$. 
Chapter 7

DISCUSSION AND CONCLUSIONS

7.1 Hypotheses Revisited

Study 1

Hypothesis 1

*The proposed regression model will yield statistically similar estimates of the true inertial weight of the upper body calculated throughout a given lifting/lowering task.*

Based on the results of Study 1, a significant amount of the variance attributable to the true inertial weight of the upper body was explained within the developed regression equation ($R^2 = 0.92$, $p < 0.0001$). Therefore, Hypothesis 1 of Study 1 is accepted.

Study 2

Hypothesis 1

*The simplified regression model developed in Study 1 will yield statistically similar, cumulative L4/L5 joint compression and anterior/posterior joint shear forces to those obtained from a dynamic rigid link biomechanical model.*

Based on the results of Study 2, this hypothesis cannot be accepted. While the Video method was shown to be statistically similar to the GOBER model overall for measures of cumulative joint compression ($p = 0.265$), magnitudes from the Video model
were statistically different in terms of cumulative anterior and posterior joint shear ($p < 0.001$, $p < 0.01$, respectively).

Hypothesis 2

*Real-time measures of cumulative spine loading created with an electromagnetic tracking device and a simplified regression-based model will be statistically similar to those produced by a dynamic rigid link biomechanical model.*

From the results of Study 2, the hypothesis cannot be accepted. Real-time measures of cumulative load from the Fastrak™ method were statistically different than the GOBER model for compression and anterior joint shear ($p < 0.05$ for both).

Hypothesis 3

*In the event of significant differences between methods, the differences between both the real-time and video approaches will be due to interaction effects attributable to either load or lift speed magnitude, or a combination of both.*

From the results of Study 2, the hypothesis is accepted. The presence of several interaction effects within the statistical analyses if Study 2 indicated that levels of both Load and Lift Speed magnitude modulated the similarity, or accuracy of the developed methods in the prediction of cumulative low back loads.

**7.2 Discussion**

The purpose of this study was to develop a lab-based methodology that could estimate 2-D dynamic cumulative spine loading in real-time. This process required the development of a simplified biomechanical model which utilized a reduced amount of coordinate data (as opposed to a traditional biomechanical model), as well as employ the
use of an electromagnetic tracking device. To statistically validate the developed model, measures obtained via the simplified approach were compared against a dynamic biomechanical model (GOBER). Theoretically, there were two sources of error that could have caused real-time estimates of cumulative load to be significantly different than the GOBER model, that being errors inherent to the simplified model, and measurement errors within the co-ordinate data obtained by the electromagnetic tracking device. In order to separate and quantify the two error sources, estimates of cumulative load were also derived from the simplified model using video-based co-ordinate data. The results of the study indicated that there were sources of error within both the developed model and between the Fastrak™ device and co-ordinate data obtained via video digitization. They are discussed in the next 2 sections.

7.2.1 Use of the Simplified Model

From a comparison of the measurement similarities between the GOBER model and the Video method, it was identified that statistically, magnitudes from the 2 methods were similar only in measures of cumulative compression. Furthermore, the interaction effect between Load and Method indicated that as the mass of the load was increased, so too was the difference in magnitude between methods. To re-iterate, the simplified model used within the Video method used two regression equations to solve for the moment arms of the upper body centre of mass and the hand held load for any given posture. As was observed in the results, an increase in load resulted in increased in relative error between the simplified Video method, and the criterion measure, the GOBER model.
It is quite possible that this increase in error is a function of the $MA_{LD}$ regression equation, which was approximately 98% accurate to the true moment arm measure, with an RMS error of 0.03 m. By increasing the load in the hands, the relative contribution of moment due to the load towards the total moment acting on L4/L5 becomes more significant. Therefore, any measurement error associated with the $MA_{LD}$ regression equation becomes greater, resulting in larger error between the GOBER model and the simplified approach. It is apparent that this measurement error is also associated with the differences between the Video method and GOBER in terms of cumulative anterior joint shear measurement, as the total joint shear is calculated by the sum of the reaction and muscle shear forces, of which the muscle shear component is derived as a percentage of total joint compression. Thus, error from within the $MA_{LD}$ regression equation is reflected in calculations of cumulative anterior joint shear as well.

Theoretically, measurement error within the Video method due to the $Inertial_{WTcm}$ equation would have been identified within lifting speed conditions. While the effect of this equation was not identified to be significant within magnitudes of cumulative compression, it was identified within measures of cumulative anterior and posterior joint shear. The best evaluation of the overall error contributions of the $Inertial_{WTcm}$ equation comes from measures of cumulative posterior joint shear, where the equation is used independently of the other regression equations. The results of the posterior shear data show that the Video method consistently underestimated the GOBER method, due to the $Inertial_{WTcm}$ equation and assumptions on which the method is based.

To conclude, it is apparent that a combination of the predictive strength of the developed regression equations, and the assumptions based within the equations used in
the Video method caused significantly different measures than the GOBER model in terms of cumulative compression, anterior joint shear, and posterior joint shear.

7.2.2 Use of the Fastrak™ Device for Real-Time Measurements

The significant differences that were identified between the Video method and the Fastrak™ method indicate that differences (errors) existed between the video digitized and Fastrak™ obtained measurements of kinematic data. Prior to each data collection, measurements were made using the Fastrak™ device for a set of known 2-D co-ordinates within the lifting area. As stated earlier, the co-ordinates obtained were always within the manufacturer’s specifications for accuracy. However, it must be noted that these are specifications for static measurements, and that none were presented for dynamic motion of the sensors. Therefore, it is possible that while the device appeared to be accurate when measures were obtained statically, increased errors were present within measures that were made by the tracking system while the sensors were moving throughout tasks. Furthermore, it is conceivable that the differences between data acquisition methods could be due to error within the video-based measures despite the calibration procedure used. Finally, it is also possible that data frames obtained from both devices were not perfectly synchronized, such that within each data frame, one device could have derived measurements at the beginning of a data frame, and the other at the end. As a result, a maximum of 0.33 seconds could have existed between recordings made by the devices, and thus it was possible that the postures analyzed by both methods were not actually the same, particularly in tasks with a high lifting speed. However, for the most part, the real-time Fastrak™ estimates were statistically similar when compared to the Video
model, except in the measurement of anterior joint shear, where the differences between measurement devices was most apparent, as a significant difference was observed between the Video Model and the Fastrak™ model for each condition level, and overall.

Despite the subtle differences between data obtained via the Fastrak™ and through video digitization, the real-time capabilities of the Fastrak™ were noted to be extremely advantageous in this study. For example, for each subject, using the video-based approach, approximately 2.5 – 3 hours was required for data acquisition, digitization, and processing time within GOBER to yield cumulative loads for all tasks. Conversely, using the Fastrak™, cumulative load data were obtained following the completion of each task. Therefore, while 2.5 – 3 hours were required for processing of video data, the Fastrak™ data were completely processed and stored by the end of a given collection session, which usually lasted about 25 minutes.

7.2.3 Functional Relevance of the Results

Based on the ANOVA results, it would seem as though the results of the study indicated that the developed model, through both video and electromagnetically obtained methods, should not be used toward future cumulative load research. However, it can be argued that the functional relevance of the differences found within the ANOVA design are minimal, and that perhaps the use of an ANOVA design for application in a study such as this, is statistically speaking, too strict of a measurement tool as an ANOVA test is fundamentally designed to find differences between groups, and not to prove similarities. Therefore, the true performance and evaluation of the developed methodologies should be made with respect to the overall relative errors, as well as the
instantaneous and cumulative accuracies (RMS error results) that were observed. This type of evaluation of a methodology has been used extensively in previous cumulative load research (Agnew et al., 2002; Callaghan et al., 2001; Keown et al., 2001).

In terms of relative error, the Video method was shown to have overall errors of -1.67% ± 5.91 %, 10.05% ± 7.29%, and -15.11 ± 22.20% for measures of cumulative compression, anterior joint shear, and posterior joint shear, respectively, with an average RMS error of 167.91 N ± 80.68 N, 25.58 N ± 9.73 N, and 9.38 N ± 3.72 N and overall cumulative RMS errors of 0.57 kN·s, 32.3 N·s, and 58.62 N·s. Similarly, the real-time Fastrak™ method had relative errors of -3.40% ± 3.57%, 4.88% ± 6.47%, and -10.34% ± 33.27% for the same measures, when compared to GOBER, and RMS errors of 154.83 N ± 78.97 N, 23.63 N ± 8.48 N, and 9.73 ± 3.98 N, respectively, and cumulative RMS errors of 0.636 kN·s, 34.92 N·s, and 53.73 N·s. Although these results are highly variable, particularly with respect to the shear measures, the overall mean errors are similar to relative errors that have been reported within alternate means of data reduction for cumulative load research, including reduced rate video sampling (Andrews et al., 2003), use of quasi-dynamic modeling (Keown et al., 2001), and posture matching techniques (Callaghan et al., 2003), which have all been reported to have acceptable relative errors that are at or below 10%.

With respect to RMS error, the instantaneous RMS errors noted within the developed methods are quite similar to those reported by Potvin (1997), who used a similar modeling approach (reported RMS errors between 147 – 183 N). It must be noted however, that the results reported by Potvin (1997) were statically based. Therefore, the developed model has extended on the work of Potvin (1997) as the results of this study
indicate that the use of NIOSH equation measurements can be used to predict dynamic estimates of lumbar compression as well, with accuracy similar to that observed for static modeling.

In regards to use of the Fastrak™ device, the observed relative errors in measurement of dynamic cumulative compression were similar to relative errors reported by Agnew et al. (2002) who used the device to obtain static cumulative compression estimates. Furthermore, it must be noted that while the developed model performs comparably with alternate means in the literature that were designed to reduce the difficulty associated with cumulative load estimation, the methods developed via this investigation are dynamic, as opposed to static measures. The developed model therefore builds on the existing literature, as both the real-time and video-based applications of this model provide a simplified means of estimating cumulative load, while at the same time retaining the inertial effects attributable to both the upper body, and any hand held load. Therefore, while the results of the study have identified significant differences between the developed methods and the criterion measure (GOBER), through an examination of the relative and RMS errors observed within the study, it can be concluded that the use of the developed model, through either video-based or real-time methods, is valid for further use in the examination of cumulative loads incurred during lifting and lowering tasks.

7.2.4 Limitations of the Study and Developed Methods

With regard to the above statement, it must be noted that there are limitations to the use of the model, and use of the Fastrak™ method for real-time documentation of cumulative loads. They are as follows:
1. The methods can only be used for lifting and lowering tasks.

2. Until further testing, the methods are only valid for loads ranging from 0 to 12.5 kg, as that is the load range that they were developed with.

3. Although dynamic, both methods are constrained to lifting/lowering in the sagittal plane.

4. The simplified model and methods were developed and validated with a lifting/lowering task that, averaged across subjects, was from floor to shoulder height and vice-versa. Therefore, the validity of these methods for lifts outside of this volume is unknown.

5. Use of a Fastrak™ device is confined to a controlled environment, like a laboratory, without the presence of large metallic objects, or sources of electromagnetic noise.

7.3 Conclusions

While research interest related to the dangers of cumulative loading in the low back has continued to develop, it is apparent that there are several logistical issues related to the monitoring and measurement of this potential cause of pain and/or injury. Primarily, the most difficult component related to cumulative load research is the acquisition and processing of data. The purpose of this study was to develop an alternate means of measuring cumulative loads that would in turn, reduce the processing demands of this type of research. The results of the current study are consistent and comparable with alternate forms of data reduction in terms of accuracy, while at the same time, quantifying loads using a dynamic, as opposed to a static model. However, the real-time
nature of the presented method is quite advantageous over other methods, such as posture sampling, or reduced rate digitization, due to the amount of time saved through acquisition of data in real-time. In addition, the developed model can also be used with video-based data, and has distinct advantages due to the reduced amount of marker coordinate data required, thus reducing the time and difficulty associated with the digitization process.

In closing, the results of this study offer yet another alternate, efficient means of measuring cumulative loading in the low back. The use of this model is attractive for implementation in the lab, or in industry by ergonomists, as the model uses measurements that are common to all ergonomists, some of whom might not be familiar with full body dynamic biomechanical modeling. Furthermore, the developed methods expand on the existing literature in that the developed model measures cumulative loads dynamically, as opposed to statically, while at the same time, reducing the processing demands commonly associated with cumulative load research.

7.4 Suggestions for Future Research

Based on the results of the study, there are several suggestions for future research. From the developed simplified model it is evident that within 2-D tasks, the use of two moment arms to represent the upper body mass and the load can determine measures of peak and cumulative load. Use of this approach greatly reduces the amount of coordinate data required to yield force estimates at L4/L5, thereby making the collection and digitization of video-based data easier. While the regression models used within this study explained a significant amount of variance of their respected measures, they could
be improved, possibly through the use of different variables, or measurement sites, such as replacing the ankle marker measurement with a measurement of the internal location of L4/L5. By doing so it is possible that more accurate moment arm predictions could be made using this joint marker instead of the ankle, and that subsequent measurement of peak and cumulative forces would be more precise.

In addition, future research should focus on the use of the Fastrak™ device, for both the real-time and 3-D capabilities the device offers. One of the major limitations of this study was the fact that the developed methods are constrained to 2-D lifting in the sagittal plane. While the Fastrak™ was accurate in estimating cumulative loads for these tasks, further research should be conducted to document the accuracy of dynamic 3-D estimates of cumulative load that could be made using the Fastrak™ device. Validation of such a method would facilitate the real-time collection of cumulative loads associated with tasks other than lifting, such as non-occupational tasks. If valid, use of the Fastrak™ for dynamic 3-D documentation of cumulative loading would yield more detailed estimates about the magnitudes, directions and rates of loads occurring in the low back, while at the same time, keeping the processing demands minimal.
REFERENCES


APPENDIX

Ethics Clearance and Research Information Forms
RESEARCH INFORMATION FORM

Measuring 2-D Dynamic Cumulative Spine Loading in Real Time

You are asked to participate in a research study conducted by: Mike Agnew and Dr. David Andrews at the University of Windsor

If you have any questions or concerns about the research, please feel to contact
Mike Agnew, Graduate Student, Faculty of Human Kinetics, University of Windsor (253-3000 x2468; Room 221 HK Building; mj_agnew@hotmail.com) or home: 255-0191.

PURPOSE OF STUDY

The purpose of this study is to develop and validate a real time method for measuring cumulative spine loading during manual materials handling tasks. Current methods employed to measure cumulative spine loading are costly and time consuming to the researcher. The development of alternate measurement techniques, such as the one investigated in this study, offers increased efficiency towards future measurement of cumulative spine loading in ergonomics research.

PROCEDURES

Subjects will be required to perform a series of lifting and lowering tasks that are varied by load, lifts style, and lift speed. Subjects are required to wear reflective joint markers for the purposes of automatic digitization into biomechanical computer software. The joint markers will be attached to the skin using medical adhesive at the following locations; hand, wrist, elbow, shoulder, ear canal, C7/T1 intervertebral joint, L4/L5 intervertebral joint, and the ankle. Sensors from an electromagnetic tracking device will also be attached at the hand, C7/T1 intervertebral joint, L4/L5 intervertebral joint, and the ankle.

POTENTIAL RISKS AND DISCOMFORTS

The conditions and trials will occur within a fairly short time frame, and participants may experience some mild fatigue in the upper and lower back, shoulders and upper arms. Muscle stiffness may result after the collection, but this should be no more than would be experienced after any unaccustomed physical activity. If you have any concerns about a pre-existing back injury, please do not participate. In addition, the attachment the reflective markers and sensors may cause a slight skin irritation. This discolouration will not last longer than a day or two and poses no health risk to the participant.

POTENTIAL BENEFITS TO SUBJECTS AND/OR TO SOCIETY

Subjects will benefit from this study by gaining insight into Kinesiology research, in particular, biomechanical modelling and research dedicated to cumulative spine loading.

PAYMENT FOR PARTICIPATION

Participants will be recruited on a volunteer basis. There will be no compensation for participation in the study.
CONFIDENTIALITY

Any information that is obtained in connection with this study and that can be identified with you will remain confidential and will be disclosed only with your permission. Only the researchers mentioned above will know your identity and personal information. This information will be stored in a secure computer in the ergonomics laboratory and will not be discussed or displayed in any form that would provide an indication of your identity.

PARTICIPATION AND WITHDRAWAL

You can choose whether to be in this study or not. If you volunteer to be in this study, you may withdraw at any time without consequences of any kind. You may exercise the option of removing your data from the study. You may also refuse to answer any questions you don’t want to answer and still remain in the study. The investigator may withdraw you from this research if circumstances arise which warrant doing so.

RIGHTS OF RESEARCH SUBJECTS

You may withdraw your consent at any time and discontinue participation without penalty. This study has been reviewed and received ethics clearance through the University of Windsor Research Ethics Board. If you have questions regarding your rights as a research subject, contact:

Research Ethics Co-ordinator
University of Windsor
Windsor, Ontario
N9B 3P4

Telephone: 519-253-3000, # 3916
E-mail:
CONSENT TO PARTICIPATE IN RESEARCH

SIGNATURE OF RESEARCH SUBJECT

I understand the information provided for the study "Measuring 2-D Dynamic Cumulative Spine Loading in Real Time" as described herein. My questions have been answered to my satisfaction, and I agree to participate in this study. I have been given a copy of this form.

Name of Subject

Signature of Subject

Date

SIGNATURE OF INVESTIGATOR

In my judgement, the subject is voluntarily and knowingly giving informed consent to participate in this research study.

Signature of Investigator

Date
VITA AUCTORIS

Name: Michael James Agnew

Place of Birth: Hamilton, Ontario, Canada

Year of Birth: 1977

Education:
- Brantford Collegiate Institute and Vocational School
  1991 – 1996

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  1996 – 2000 Honours BA

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