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EVALUATION OF THE TRUNK BIOMECHANICAL RESPONSES TO SEATED WHOLE-BODY VIBRATION EXPOSURE

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Cette thèse intitulée :

EVALUATION OF THE TRUNK BIOMECHANICAL RESPONSES TO SEATED WHOLE-BODY VIBRATION EXPOSURE

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RÉSUMÉ

L'exposition des travailleurs aux vibrations globales du corps est reconnue comme étant un facteur de risque dans l'étiologie des maux de dos. Par contre, les mécanismes selon lesquels les vibrations globales du corps peuvent causer des blessures sont inconnus. Cette thèse propose le postulat que les vibrations globales du corps peuvent affecter la stabilité de la colonne vertébrale. Utilisant comme cadre conceptuel le modèle de Panjabi portant sur la stabilité de la colonne, plusieurs mesures biomécaniques ont été utilisées pour examiner les effets des vibrations globales du corps sur les trois sous-systèmes de ce modèle, soit les sous-systèmes actif, passif et de contrôle neuromusculaire. Cette recherche a pour but d'examiner: 1) la fidélité des différentes mesures biomécaniques; et 2) les effets des vibrations globales du corps sur les trois sous-systèmes par le biais des mesures biomécaniques suivantes: 1) les réponses réflexes des muscles du dos et l'équilibre en position debout pour évaluer le sous-système du contrôle nerveux; 2) la variation de longueur de la colonne vertébrale ou "spinal shrinkage" (par stadiométrie), pour évaluer le sous-système passif; 3) l'activation et la fatigue des muscles du dos (par électromyographie), pour évaluer le sous-système actif.

Pour atteindre ces objectifs, trois études ont été conduites séparément. Dans l'étude 1, la théorie générale de la fidélité a été utilisée comme structure pour examiner la fidélité des mesures biomécaniques chez 15 hommes en santé. Cette étude a déterminé que la majorité des variables dépendantes ont une fidélité variant de pauvre à moyenne

(ICC < 0.75). Cependant, faire la moyenne de 7 essais ou plus, essais effectués durant la même journée de tests, permet d'atteindre un niveau de fidélité acceptable. L'étude 2 a examiné quelles mesures biomécaniques et quelles variables étaient le plus susceptibles d'être sensibles à l'exposition à des vibrations globales du corps. Douze hommes en santé ont été exposés à 60 minutes de vibrations verticales alors qu'ils étaient assis sur un siège rigide en métal et qui ne comportait pas de soutien au dos. Les caractéristiques de l'exposition aux vibrations étaient tout juste sous la norme permise afin de simuler des conditions "extrêmes". Une condition sans vibration a aussi été réalisée. Les résultats ont démontré une plus grande activité musculaire pour certains muscles lors de la vibration. Cependant, les mesures biomécaniques d'équilibre, de réponses réflexes et de "spinal shrinkage" sont demeurées les mêmes et aucune présence de fatigue musculaire n'a été observée. L'étude 3 a examiné si l'exposition aux conditions vibratoires typiques trouvées dans les véhicules miniers (caractéristiques vibratoires comparables et posture du tronc asymétrique). Comme dans l'étude 2, il n'y a pas eu aucun effet sur les réponses réflexes et sur l'équilibre suivant 60 minutes d'exposition. Pour certains muscles, l'activité musculaire était plus grande durant les vibrations que lors de l'absence de vibrations. Une évidence de fatigue des muscles du dos a été observée dans les deux conditions (avec et sans vibration), suggérant que la posture asymétrique seule en soit la cause.

Ces résultats suggèrent que les mécanismes qui conduisent à des blessures au bas du dos ne seraient pas liés à la stabilité de la colonne vertébrale. Cependant, considérant les études de la dernière année sur le sujet, cette possibilité doit être réévaluée dans d'autres études comportant des méthodologies et des mesures plus raffinées.

ABSTRACT

Occupational whole-body vibration (WBV) has been shown to be associated with low-back disorders. However, the mechanisms by which WBV can cause these disorders is unknown. It is proposed in this thesis that one potential mechanism is that WBV effects spinal stability. Using Panjabi's model of spinal stability as the conceptual framework, several biomechanical measures were used to investigate the effects of WBV on the three subsystems of the model: active, passive and neuromuscular control. The objectives of this research were to investigate the: 1) reliability of the different biomechanical measures; and 2) effects of WBV on the three subsystems. The biomechanical measures that were used included: 1) back muscle reflex response and standing balance, to assess the neural control subsystem; 2) spinal shrinkage (through stadiometry), to assess the passive subsystem; and 3) back muscle activation and fatigue (through electromyography), to assess the active subsystem.

To reach the objectives, three separate studies were conducted. In Study 1, the Generalizability Theory was used as the framework to investigate the reliability of the biomechanical measures in 15 healthy males. This study found that the majority of the dependent variables displayed poor to moderate reliability (ICC < 0.75) and that averaging of several trials is shown as a practical strategy for improving reliability. For the majority of the variables, acceptable reliability could be achieved when at least 7 or more trials are averaged during the same testing day. Study 2 investigated which biomechanical measures and variables were most likely to be sensitive in detecting

responses due to WBV exposure. Twelve healthy males were exposed to 60 minutes of vertical vibration while seated on a rigid, metal seat with no backrest. The effect of WBV under "extreme" conditions (i.e., magnitude of vibration exposure near the resonant frequency, rigid seat), while still below the limits of exposure set by the International Organization for Standardization. A no-vibration (control) condition was also carried out. Depending on the muscle, muscular activity was higher during the vibration condition than the no-vibration condition. However, the other biomechanical measures (back muscle reflexes, balance, spinal shrinkage) were not sensitive to WBV and no presence of muscular fatigue was found. Study 3 explored whether exposure to conditions typically found in a large load-haul dump mining vehicle (comparable vibration spectral signature and trunk asymmetrical posture) have an effect on the biomechanical responses to seated WBV. As in the second study, there was no effect on the reflex response and balance following 60 minutes of exposure and for certain muscles muscular activity was higher during the vibration condition than the no-vibration condition. Back muscle fatigue was substantiated for all muscles in both vibration and no-vibration conditions indicating that the trunk asymmetrical posture itself induced this effect.

In summary, there was no significant effect of seated WBV exposure on any of the measures that investigated the three subsystems that control spinal stability. These findings suggest that the mechanisms that lead to low-back pain and disorders may not be related to deficits in spinal stability. However, considering recent findings in other laboratories, the possibility that WBV effects spinal stability remains. Future studies with different experimental measures and using more sensitive and reliable outcome measures should be conducted to clarify this situation.

CONDENSÉ EN FRANÇAIS

L'exposition des travailleurs aux vibrations globales du corps est reconnue comme étant un facteur de risque dans l'étiologie des maux de dos. Bien qu'il n'y ait pas de relation exposition-réponse clairement définie, plusieurs études épidémiologiques suggèrent qu'il existe une relation nette entre l'exposition aux vibrations globales du corps et le développement de maux de dos (Bernard, 1997). Le rôle des vibrations dans l'incidence des affections vertébrales est toutefois difficile à démontrer. Il n'est pas possible, à partir des travaux existant, d'établir une relation de cause à effet entre les vibrations et les blessures au dos parce que les effets biomécaniques des vibrations sur les sujets humains sont peu documentés et les mécanismes selon lesquels les vibrations globales du corps peuvent causer des blessures sont inconnus.

Cette thèse propose le postulat que les vibrations globales du corps peuvent affecter la stabilité de la colonne vertébrale. D'un point de vue clinique, la stabilité est définie par l'habileté de la colonne vertébrale à limiter, sous l'effet de chargements physiologiques, les déplacements articulaires de façon à ne pas endommager ou irriter les différents tissus de la colonne. Selon Panjabi (1992), la stabilité de la colonne est contrôlée par trois sous-systèmes qui doivent être considérés en concert, soit les sous-systèmes passif (ex: disques intervertébraux), actif (ex : muscles) et de contrôle neuromusculaire (ex : système nerveux central). Si un ou plusieurs sous-systèmes ne parviennent pas à fonctionner de façon optimale, la colonne vertébrale risque de subir une lésion. Utilisant comme cadre conceptuel le modèle de Panjabi, plusieurs mesures biomécaniques ont été utilisées pour examiner les effets des vibrations globales du corps sur les trois sous-systèmes en question.

Cette recherche avait pour but d'examiner: 1) la fidélité des différentes mesures biomécaniques; et 2) les effets des vibrations globales du corps sur les trois sous-systèmes. Les mesures biomécaniques suivantes ont été considérées: 1) les réponses réflexes des muscles du dos et l'équilibre en position debout pour évaluer le sous-système de contrôle neuromusculaire; 2) la variation de longueur de la colonne vertébrale ou "spinal shrinkage" (par stadiométrie), pour évaluer le sous-système passif; 3) l'activation et la fatigue des muscles du dos (par électromyographie), pour évaluer le sous-système actif. Pour atteindre ces objectifs, trois études ont été menées séparément. La première (Étude 1) consistait à vérifier la fidélité test-retest des mesures biomécaniques utilisées lors des tests expérimentaux. La seconde (Étude 2) a examiné quelles mesures biomécaniques et quelles variables étaient les plus susceptibles d'être sensibles à l'exposition à des vibrations globales du corps. La troisième (Étude 3) était similaire à la seconde à l'exception que les sujets étaient soumis à des vibrations de même amplitude et fréquence que celles retrouvées dans un véhicule minier (chargeuse navette de grande capacité), avec le même type de siège et avec l'adoption des mêmes postures.

Les mesures biomécaniques

La réponse réflexe

La réponse musculaire réflexe a été évaluée sur un appareil spécialement conçu pour générer une perturbation soudaine des muscles du dos. Les réflexes étaient initiés lors de l'application soudaine d'une charge correspondant à 35% de la masse du tronc. Le sujet était placé dans l'appareil tel qu'illustré à la figure 4.1. La perturbation avant du sujet était transmise au moyen d'un câble parallèle au sol fixé à la hauteur de la 8^{e} vertèbre dorsale dont la tension provenait de la chute de la charge d'une hauteur de 1 cm. Le sujet était soumis à plusieurs perturbations dans lesquelles le câble était soit sans tension ou encore pré-tendu à une tension équivalente à 15% de la masse du tronc. Le signal électromyographique de trois muscles bilatéraux du dos (long dorsale au niveau L1, ilio-costale au niveau L3 et multifide au niveau L5) a été enregistré avec des électrodes de surface actives de façon à déterminer la réponse réflexe de ces muscles. Également, le signal électromyographique (EMG) de deux muscles supplémentaires, c'est-à-dire le droit antérieur des abdominaux et les obliques externes, a servi de rétroaction au sujet afin de minimiser le niveau de co-contraction musculaire avant d'effectuer la perturbation. La latence de la réponse réflexe était définie par la période de temps entre le début du déplacement du tronc et le début de la réponse réflexe. L'amplitude du signal fut évaluée en calculant le ratio du signal moyen EMG pendant la perturbation sur celui du signal au repos. Les variables cinématiques, c'est-à-dire le déplacement, la vélocité et l'accélération angulaire du tronc ont été mesurées au moyen d'un potentiomètre.

Équilibre (en position debout)

L'équilibre était estimé au moyen d'une plate-forme de force sur laquelle le sujet devait se tenir debout, sur une durée de 60 s. Entre chaque essai, le participant devait se retirer de la plate-forme pour immédiatement se repositionner sur celle-ci pour l'essai suivant. La trajectoire du centre de pression a ainsi été mesurée et les principales mesures résumant le comportement de cette trajectoire ont été calculées (Prieto et al. 1996).

Stadiométrie

Un stadiomètre (Figure 6.1) a été utilisé pour évaluer la variation de longueur de la colonne vertébrale due à l'affaissement des disques intervertébraux. Le stadiomètre est une charpente métallique inclinée de 15° vers l'arrière offrant différents appuis standards aux sujets afin de pouvoir reproduire la même posture lors d'une série de mesures. Pour ce faire, des capteurs de pression sont placés le long de l'appareil de manière à fournir aux sujets une rétroaction sur leur posture. Un inclinomètre fixé sur la tête permet également de contrôler l'inclinaison de la tête. La rétroaction au sujet se réalisait au moyen d'un écran visuel (placé en face de celui-ci) qui lui indiquait si la posture adoptée se conformait exactement à celle qu'il avait précédemment choisie. La taille était mesurée à partir de la base des pieds jusqu'à la vertèbre C7.

L'activation musculaire

L'amplitude RMS (Root Mean Square, fenêtres de 0,125 s) des signaux collectés durant la période de 60 minutes en position assise ont été normalisés à une valeur de référence correspondant à l'EMG maximal obtenu lors de contractions maximales volontaires (CMVs). La fonction de la distribution des probabilités des amplitudes du signal (Amplitude Probability Distribution Function ou APDF (Jonsson, 1978)) a été utilisée pour calculer les niveaux d'activité musculaire correspondant au 10^{e} (niveau statique), 50^{e} (niveau médian) et 90^{e} (niveau maximal) centile (%ile).

La fatigue musculaire

Le signal EMG enregistré lors du test de réflexe, juste avant la charge soudaine, a été utilisé pour établir la présence de la fatigue, car cette portion du test assurait une charge et une posture constantes entre les essais. Ainsi, les premières 5 secondes avant que la charge soit appliquée ont été utilisées pour obtenir la fréquence moyenne instantanée (Karlsson et Gerdle, 2001).

Étude 1: Fidélité de certaines mesures biomécanique

L'objectif de cette étude était de vérifier la fidélité de certaines mesures biomécaniques destinées à évaluer l'effet des vibrations sur des sujets, soit la réponse réflexe, l'équilibre et la stadiométrie. Quinze hommes en santé ont participé à l'étude. Les sujets devaient se présenter à deux séances expérimentales dans un délai maximal d'une semaine. Pour la réponse réflexe, une condition avec et une condition sans pré-tension ont été réalisées. Pour l'équilibre, une condition avec les yeux fermés et une avec les yeux ouverts ont été réalisées.

La théorie générale de la fidélité (Generalizability Theory) a été utilisée pour examiner la fidélité (Shavelson et Webb, 1991). Cette théorie comporte deux parties: la première est l'étude-G qui estime l'importance relative des composantes de variance jugées pertinentes pour la mesure d'intérêt, ceci à partir des résultats du devis expérimental considéré (mesures répétées à l'intérieur d'une session de mesure et sur deux jours). La deuxième partie est l'étude-D permettant de donner une estimation de la fidélité pour divers devis de recherche (stratégies de mesures) autres que l'étude-G. Deux indices de fidélité ont été calculés, soit le coefficient de corrélation intra-classe (CCIC) et l'erreur standard de mesure (ESM).

Cette étude a démontré que la majorité des variables dépendantes ont une fidélité variant de faible à moyenne (CCIC < 0,75). Pour la réponse réflexe, dépendamment de la variable, les CCIC variaient entre 0 et 0,62. Pour l'équilibre, les CCIC variaient entre 0,02 et 0,76. Le CCIC pour le stadiométrie était de 0,3. Cependant, faire la moyenne de 7 essais ou plus durant la même journée de tests, permettait d'atteindre un niveau de fidélité acceptable.

Étude 2: Sensibilité des mesures biomécaniques

L'objectif de cette étude était d'examiner quelles mesures biomécaniques et quelles variables étaient le plus susceptibles d'être sensibles à l'exposition à des vibrations globales du corps. Pour ce faire, les caractéristiques de l'exposition aux vibrations étaient tout juste sous la norme permise afin de simuler des conditions "extrêmes", l'idée étant de maximiser les chances de détecter des effets.

Douze hommes en santé ont participé à l'expérience. Les sujets étaient soumis à une exposition de 60 minutes de vibrations verticales alors qu'ils étaient assis sur un siège rigide en métal qui ne comportait pas de soutien au dos, et les bras appuyés sur les cuisses. Le spectre en fréquence des vibrations couvrait la plage des fréquences entre 0,5 et 20 Hz, et la fréquence centrale à laquelle l'accélération maximale était produite se situait autour de 4,5 Hz, tandis que l'accélération efficace pondérée correspondante était de 1,4 ms^{-2}. Pour la condition de contrôle, le sujet devait demeurer en position assise sur le banc rigide pendant 60 minutes.

La variation de longueur de la colonne vertébrale, l'équilibre ainsi que les mesures propres aux muscles du dos (réponses réflexes, niveau d'activité, fatigue) ont été utilisées pour quantifier la réponse biomécanique aux vibrations. Les participants ont dû se présenter à six séances expérimentales, c'est-à-dire à trois reprises pour chacune des techniques de mesure (certaines permettaient d'être combinées) et sous deux conditions, soit avec et sans exposition aux vibrations. Chaque séance se réalisait au cours d'une journée différente et l'ordre des conditions expérimentales était contrebalancé à travers les sujets. Des mesures biomécaniques se faisaient immédiatement avant (PRE) et après (POST) les 60 minutes d'exposition et également lors de la période de récupération, soit à 20 (RECOV20), 40 (RECOV40), et 60 (RECOV60) minutes après l'exposition aux vibrations.

Pour les mesures d'équilibre et des réponses réflexes (temps de latence et ratio de l'amplitude EMG), la moyenne des essais a été calculée pour chacune des périodes de mesures. Cette moyenne fut alors utilisée pour une analyse de variance à mesures répétées pour deux facteurs : "condition" (vibrations vs sans vibrations) et "période de mesure" (PRE, POST, RECOV20, RECOV40, et RECOV60).

Les résultats ont démontré une plus grande activité musculaire pour certains muscles du dos lors de la vibration. Pour deux muscles, l'amplitude EMG était 33% et 71% plus élevée dans la condition avec que dans la condition sans vibration. Généralement, le 90^{e} centile de l'activité musculaire était en-dessous de 25,4% CMV. Cependant, aucun effet d'interaction *Condition* × *Temps* n'a été trouvé pour les mesures d'équilibre et de réponses réflexes et la variation de longueur de la colonne vertébrale a demeuré la même. Les valeurs de la fréquence moyenne ont augmenté dans le temps, ce qui suggère qu'aucune fatigue musculaire n'était impliquée.

Étude 3: L'effet des vibrations globales du corps sur les mesures biomécaniques

L'objectif de cette étude était d'examiner si l'exposition aux conditions vibratoires typiques trouvées dans les véhicules miniers (caractéristiques vibratoires comparables et posture du tronc asymétrique) produisait des effets tel que mesuré avec les mesures décrites ci-haut.

Douze hommes en santé ont été soumis à une exposition de 60 minutes de vibrations verticales de type aléatoire au moyen d'un simulateur reproduisant un poste de conduite sur chargeuse navette. La classe spectrale simulée variait entre 0,5 et 15 Hz avec une valeur maximale de fréquence se situant à 2,7 Hz. L'accélération efficace pondérée correspondante était de 0,86 ms^{-2}. Un siège de conduite a aussi été utilisé pour reproduire la condition type retrouvée dans les chargeuses navettes. Les postures typiquement adoptées par les conducteurs de ces types de véhicules ont aussi été simulées selon une certaine séquence, ce qui impliquait, entre autres, des postures en torsion du tronc.

Les participants ont dû se présenter à quatre séances expérimentales, c'est-à-dire à deux reprises en fonction des techniques de mesure considérées (1. équilibre; 2. réponses réflexes/fatigue/activité musculaire) et sous deux conditions, soit avec exposition aux vibrations et sans exposition. Chaque séance se réalisait sur une journée différente et l'ordre des conditions expérimentales était contrebalancé à travers les sujets. Les mesures biomécaniques se faisaient aux mêmes périodes que dans l'étude 2 (avant et après l'exposition et lors de la période de récupération) et les tests statistiques étaient donc aussi identiques.

Comme dans l'étude 2, pour certains muscles, l'activité musculaire était plus grande durant les vibrations que lors de l'absence de vibrations. L'amplitude EMG était (dépendamment du muscle) entre 22% et 48% plus élevée dans la condition avec que dans la condition sans vibration, et le 90^{e} centile de l'activité musculaire était en dessous de 15,1% CMV. Il n'y a pas eu d'effet sur les réponses réflexes et sur l'équilibre suivant les 60 minutes d'exposition (aucun effet d'interaction *Condition* × *Temps* n'a été trouvé). Une évidence de fatigue des muscles du dos a été observée dans les deux conditions (avec et sans vibration), suggérant que la posture asymétrique seule en était la cause.

Discussion et recommandations

Par rapport au premier objectif de cette recherche, c'est-à-dire d'examiner la fidélité de différentes mesures biomécaniques, la fidélité de la plupart des variables variait de faible à moyenne. L'utilisation de la moyenne de plusieurs essais s'avère donc nécessaire pour obtenir une fidélité acceptable de ces mesures. Par rapport au deuxième objectif, c'est-à-dire d'examiner les effets des vibrations globales du corps sur les trois sous-systèmes, les résultats des études 2 et 3 suggèrent que les vibrations ne sont pas responsables des effets qui furent observés. Même si les caractéristiques des vibrations, du siège et des postures étaient différentes dans les deux études, les vibrations n'ont pas démontré d'effet et ce malgré toutes les précautions méthodologiques mises de l'avant, notamment pour assurer la fidélité des mesures.

D'autres groupes de recherche dont les résultats ont paru tout récemment (2008) et aussi intéressés à l'étude de l'effet des vibrations globales du corps sur différents sous-systèmes ont observé des effets des vibrations. Un groupe a trouvé un effet sur la réponse réflexe et un autre groupe a trouvé un effet sur l'équilibre postural, ce dernier utilisant une technique de mesure plus spécifique à la région lombaire.

Concernant l'étude sur la réponse réflexe, le groupe a utilisé un siège rigide et les sujets ont été exposés à une vibration de courte durée à un niveau d'accélération de 0.22 ms^{-2}. Par contre, les valeurs réflexes latentes qui ont été rapportées étaient de > 150 ms, ce qui indique que les réflexes étaient volontaires plutôt qu'involontaires. Concernant l'étude sur l'équilibre postural, même si la durée de l'exposition était plus courte et que le niveau d'accélération était différent (1.15 ms $^{-2}$) de cette présente étude, les mesures de l'équilibre postural ont pu être enregistrées immédiatement après l'exposition, car le siège utilisé était aussi l'instrument de mesure. Cependant, il existe des limites à cette étude (voir paragraphe plus bas) qui ont été expliquées dans cette présente étude.

La présente étude n'a pas mené à des résultats positifs, cela a eu un rigueur qui n'a pas été démontré dans les études récentes. Cette recherche comprenait une condition sans vibration (groupe de contrôle), ceci présenté dans un ordre contrabalancé. Dans l'étude 3, des conditions plus réelles de travail ont été utilisées (ex: les caractéristiques vibratoires, le siège, les postures adoptées). La durée d'exposition dans la posture assise était aussi plus longue que dans les autres études.

Les résultats de cette recherche suggèrent que les mécanismes qui conduisent à des blessures au bas du dos ne seraient pas liés à la stabilité de la colonne vertébrale. Cependant, considérant les études parues dans la dernière année sur le sujet (autres groupes de recherche), cette possibilité doit être réévaluée dans d'autres études comportant des méthodologies et des mesures plus raffinées.

Les études futures devraient étudier différentes caractéristiques de vibrations appliquées dans les différentes directions (pas seulement verticales), ce qui représenterait des conditions de travail plus réalistes. D'autres techniques de mesure qui ont été utilisées dans des études plus récentes pourraient également être employées. Notamment, des mesures qui sont plus sensibles à détecter la fatigue musculaire lors de contractions de bas niveau devraient être employées. En considérant les variables à ne pas négliger (ex : vibrations dans d'autres directions, postures, le siège), avec un devis expérimental adéquat et avec des mesures qui ont le meilleur potentiel pour détecter les effets, une meilleure compréhension de la relation entre les vibrations globales du corps et les maux de dos deviendra possible.

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LIST OF ABBREVIATIONS

Abbreviation	Long form
ANOVA	Analysis of variance
AP	Anterior-posterior
APDF	Amplitude probability distribution function
COP	Centre-of-pressure
D-Study	Decision study
EMG	Electromyography
G-Study	Generalizability study
ICC	Intra-class correlation coefficient
LONG	Longissimus muscle
ML	Medial-lateral
MULT	Multifidus muscle
SEM	Standard error of measurement
WBV	Whole-body vibration

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INTRODUCTION

Back injury and low back pain (LBP) are serious public health issues associated with considerable disability, healthcare use, and societal costs. In 2003, the World Health Organization (WHO) deemed low back pain (LBP) a severe public health issue and the leading cause of chronic health problems and long-term disability (Ehrlich, 2003). Further, LBP accounts for approximately 37% of occupational injury globally (Punnett et al., 2005). It accounts for approximately between 20% and 30% of all workers' compensation claims and up to 50% of all direct compensation costs (Kerr et al., 2001). Conservative estimates of annual expenditures in the United States of the direct costs of LBP range between 13 and 20 billion dollars (Bernard, 1997). The total costs are estimated to be much higher due to indirect costs such as production loss, hiring and training of replacement workers, overtime, and administrative costs.

With the enormous amount of money being lost and pain being suffered due to this problem, and given that a large proportion of the adult population will experience an episode of back pain and injury at some point in their lives, even small advances toward the understanding of lumbar spine etiology will have an impact. Therefore, it is essential that a better understanding of the mechanisms of LBP and injury be attained. The understanding of injury mechanisms is prerequisite to the development of effective prevention and rehabilitation methods. The knowledge as to the mechanisms that may lead to back injury is still incomplete, and the physiological and biomechanical effects associated with exposure encountered in true working conditions is a research area that is just starting to be developed.

A large body of evidence supports the importance of physical load factors in the development of LBP (Nachemson & Jonsson, 2000; National Research Council, 2001). Among physical exposures encountered in working conditions, there is moderate to strong evidence concerning the risk of LBP and spinal disorders associated with exposure to whole-body vibration (WBV) (Bernard, 1997; Bovenzi & Hulshof, 1999; Hoogendoorn, van Poppel, Bongers, Koes & Bouter, 1999; Leboeuf-Yde, 2004; National Research Council, 2001; Waddell & Burton, 2000). It is estimated that in Canada, the United States and some European countries, approximately 4 to 7% of employees are exposed to potentially harmful levels of WBV (Bovenzi, 1996). Wasserman et al. (1997) claim that in the United States, approximately 7 million workers are exposed to WBV. In Québec, data from the Commission de la santé et de la sécurité du travail (CSST) indicate that between 2002 and 2004, a total of 14 757 days were compensated, costing almost 1.45 million dollars due to injuries related to occupational exposure to whole-body vibration from a vehicle or mobile equipment. Of this amount, injuries to the back accounted for 78% of the total number of injuries.

Although it has been shown that WBV is a moderate to strong risk factor in the etiology of LBP, its role in the effects on the spine, and other organs, is still inconclusive and there is no established dose-response relationship (Thalheimer, 1996). The cause of LBP and injury is multifactorial in nature, and thus makes it sometimes difficult to establish a clear relationship between these factors and the LBP problem. In occupations in which workers are exposed to seated WBV, this group is often also exposed to other risk factors for LBP. Some of these factors include prolonged sitting (Lis, Black, Korn & Nordin, 2007), awkward postures (i.e., rotated trunk and neck), manual materials handling (MMH) (i.e., pulling lifting), poor climatic conditions (i.e., extreme heat or cold), and also psychosocial factors. The interaction due to the combined exposure to two or more of WBV, posture and MMH are the main contributors of LBP (Okuribido, Magnusson & Pope, 2008).

It is often postulated that prolonged vibration exposure may cause spine pathology through mechanical damage. Studies have shown that exposure to WBV causes spine changes that may be related to LBP, which include lumbar disc flattening, increased intradiscal pressure, disc herniation, and microfractures in the vertebral endplates (Wilder & Pope, 1996). The vertebral endplate, followed by the intervertebral disc are the structures most sensitive to high WBV exposure (Wikström, Kjellberg & Landström, 1994). Vibration exposure may cause disc degeneration and fractures of the vertebral endplate, and may cause creep of the spinal motion unit. WBV exposure may also lead to back muscle fatigue (Hansson, Magnusson & Broman, 1991; Wikström, Kjellberg & Landström, 1994; Wilder, Magnusson, Fenwick & Pope, 1994).

In addition to the mechanical damage described in the previous paragraph, another hypothesis as a potential mechanism for LBP related to WBV is that WBV leads to instability of the spinal column. This instability places the spine at risk in situations such as MMH. Bone, discs, ligaments, and muscles in effect, provide lumbar spine stability. Thus, any impairment in their function may in fact, lead to instability of the lumbar spine. The stability of the spine is important in order to avoid intervertebral movements that could increase the risk of back problems. Panjabi (1992) has provided a model of spinal stability as being controlled by passive (e.g., intervertebral discs), active (e.g., muscles), and neural (e.g., central nervous system) subsystems working interdependently to maintain the stability of the spine. If one (or more) subsystem fails to function optimally, the spinal column is at greater risk for injury. An increasing body of scientific evidence supports this hypothesis (Preuss & Fung, 2005).

The conceptual framework around which this thesis will be based is that of Panjabi's (1992) lumbar-stability hypothesis. Using this hypothesis, however, implies that many injury pathways may be involved. This project will explore the use of different techniques that could potentially measure biomechanical responses that would have an impact on the different subsystems in the lumbar-stability model. Laboratory studies on these responses after WBV exposure, though not new, are scarce. Lumbar stability is very difficult to measure, and the measures used are usually indirect and their metric qualities (e.g., reliability) are not well documented. Furthermore, there is little, if any, discussion on the interaction of these different responses and the underlying sensory-motor control mechanisms that may affect trunk stability and postural control. Thus, through controlled experimental studies, this project will attempt to further the knowledge in this area.

WBV may alter the load distribution among the passive tissues of the spine. Spinal shrinkage (or height loss), measured by stadiometry, has been used as a non-invasive approximation of the spinal deformation (Eklund & Corlett, 1984; McGill et al., 1996; van Dieën & Toussaint, 1993). Intervertebral disc deformation can lead to an increased risk of injury to the disc itself, but also to other structures surrounding the spine (Adams et al., 2002).

In addition, WBV may alter some neuro-sensory functions, such as trunk reflex responses and postural balance. Delayed trunk muscle reflex responses (Cholewicki et al., 2005) as well as poor postural balance (Takala & Viikari-Juntura, 2000) could increase the risk of low back injuries. For example, this may be a possible mechanism of injury for a worker who is first exposed to a period of WBV and then immediately following, must perform MMH tasks (i.e., lifting a load) with the risk of a sudden load. With decreased lumbar stability, and if reflexes remain perturbed after a relatively long exposure to vibration, this injury pathway may apply and hence increasing the risk while performing such tasks. Furthermore, with decreased balance, there is a greater likelihood of slipping and/or falling (especially in working environments where there is uneven terrain).

Whole-body vibration may also result in muscle fatigue due to acute reflex activation of the primary muscle spindle fibres. Lumbar muscle fatigue has been shown to reduce neuromuscular control of trunk movement (Cholewicki, Polzhofer & Radebold, 2000; McGill, 2001; McGill & Cholewicki, 2001; Ng, Parnianpour, Richardson & Kippers, 2003; Parnianpour, Nordin, Khanovitz & Frankel, 1988) and thus may contribute to low back injury by compromising the stability of the spine.

Evaluation of the biomechanical responses to seated WBV exposure is important in improving our understanding of the role WBV might play in the development of LBP or injury. If reliable and objective measures exist to quantify these biomechanical response in relation to WBV, and if these responses are sensitive to WBV, strategies to minimize or optimize the biomechanical response can be sought and help improve preventing low-back disorders. The main questions of interest are:

1. Are the different biomechanical measures that could potentially be used to assess the different subsystems in Panjabi's spinal stability model reliable?
2. To what extent are these biomechanical measures sensitive to WBV and what are the effects of WBV on the three subsystems?

To help improve our understanding of the effect of WBV on neuro-control mechanisms, objective, reliable, and sensitive measures must be used to assess the different potential effects. The main purpose, therefore, of the proposed research is to develop objective, non-invasive and reliable measures to explore and quantify the effects of WBV on trunk biomechanics and balance. If any effects are found, new hypotheses will possibly be formulated and then tested in the future. The proposed research will be carried out in three different studies. Each study will address different objectives to answer the different research questions. More specifically, the objectives are:

Objective 1: To verify that the targeted biomechanical measures expected to be sensitive to WBV can be handled with confidence in the laboratory and to determine if the measures achieve similar results over repeated measures (reliability).

Objective 2: To determine the sensitivity of the different measurement techniques to WBV exposure.

Objective 3: Using the most promising (i.e., most reliable and sensitive) measurement techniques, what are the effects of WBV exposure in simulated working conditions (i.e., typical mining vehicle seat, postures, and vibration exposure) on spinal stability.

CHAPTER 1

REVIEW OF THE LITERATURE

1.1 Risk factors for low-back disorders

The aetiology of low back disorders is assumed to be of multifactorial origin. Individual factors, as well as work-related (and non work-related) physical and psychosocial factors, may play a role in its development. A conceptual model (Figure 1.1) of the influences and possible roles that various factors may have in the development of musculoskeletal disorders (National Research Council, 2001) suggests that these factors may be divided into two groups: individual characteristics and workplace (or occupational) factors. As shown in Figure 1.1, the literature relating to epidemiologic studies of low back disorders has evaluated the link between the workplace and low back disorders primarily along two dimensions: 1) exposure to external loads in the workplace and its association with low back disorder outcomes, and 2) the association between organizational factors and the social context (also called psychosocial factors).

External physical loads may cause acute effects whereby the tissue tolerance is exceeded, or it may have a chronic effect whereby the tissue tolerance gradually decreases over time to a point where previously acceptable mechanical loads result in LBP (Adams & Dolan, 1995). Furthermore, physical loads may cause different tissue responses (i.e., tissue deformation, altered metabolism, altered circulatory patterns, inflammation, muscle fatigue) meaning that LBP may results from an effect on multiple spinal structures (National Research Council, 2001). Although there are several plausible mechanisms for the relationship between physical load and low-back disorder outcomes, there is still no full understanding of the complex process of how physical factors result in physiological responses, ultimately leading to musculoskeletal symptoms.

Physical and psychosocial load at work are related since both are determined by workplace design and work organization. Psychosocial work characteristics not only can influence the physical load through changes in posture, movement, and exerted forces (Bongers, de Winter, Kompier, & Hildebrandt, 1999; Sauter & Swanson, 1996; Theorell, 1996), but they may directly trigger physiological mechanisms (i.e., increased muscle tension) or may increase psychological strain, which, in turn, may increase muscle tension or hormonal excretion.

Individual factors include factors such as age, gender, smoking habits, exercise behaviour, physical fitness and training, anthropometry and coping skills. These factors may be independent of LBP, but may also influence the relationship between physical and psychosocial factors as well as LBP.

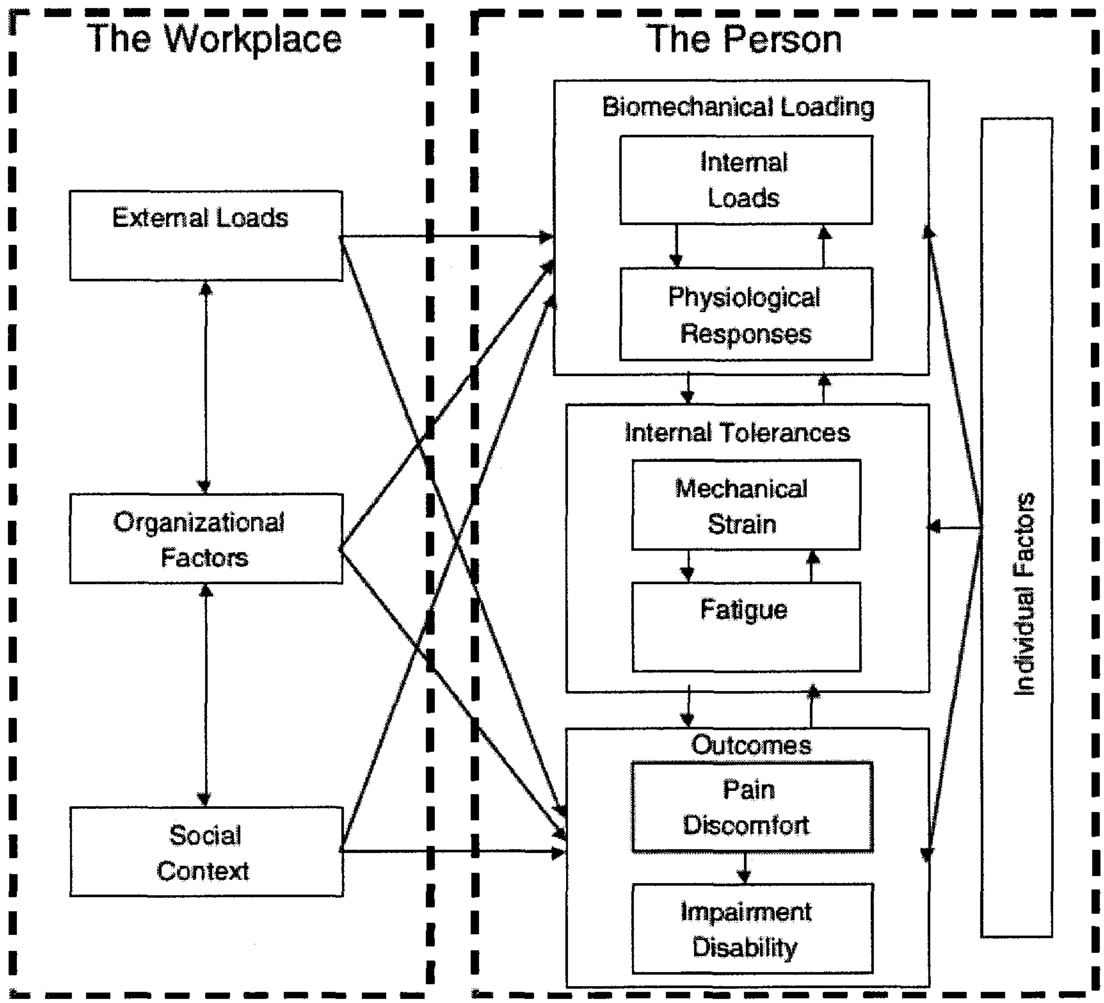

Figure 1.1: Conceptual model of the possible roles and influences that various factors may play in the development of musculoskeletal disorders (National Research Council, 2001)

1.1.1 Work-related physical risk factors

The review of the observational epidemiology literature has shown support for the linkage between external load exposure in the workplace and increased low back disorders. In the review by the National Research Council (2001), it was concluded that there is a clear relationship between low back disorders and physical load, frequency of bending and twisting, physically heavy work, and WBV (with risk ratio estimates up to 9-fold and attributable fractions of between 11 and 66 percent). In a systematic review by Hoogendoorn et al. (1999), strong evidence is found for MMH, bending and twisting and WBV, however only moderate evidence for patient handling and heavy physical work, and no evidence for standing or walking. They also found that sitting alone does not increase the risk, which is corroborated by others (Lis, Black, Korn, & Nordin, 2007; Okunribido, Magnusson, & Pope, 2008). Other physical factors proving to be strong risk factors include peak lumbar shear forces and cumulative loading (Kerr et al., 2001). Sudden loading events in the workplace have also been related to a high incidence of low-back injuries (Magnusson et al., 1996).

Whole body vibration has repeatedly been identified as a risk factor for low back pain (Bernard, 1997; Bovenzi, 1996; Lis et al., 2007; Seidel, 1993). Several authors have concluded that there is an association between low-back disorders and WBV (Bovenzi & Hulshof, 1999; Lings & Leboeuf-Yde, 2000), however, the injury mechanisms linking WBV to low back disorders needs to be better understood. The difficulty in identifying any specific injury mechanism is that in occupations in which workers are exposed to WBV, other physical risk factors are also present such as awkward postures, prolonged sitting, awkward postures and MMH (i.e., loading and unloading materials from a vehicle) (Lis et al., 2007; Okunribido et al., 2008).

1.1.2 Work-related psychosocial risk factors

An association (with risk estimates between 1 and 5) has been established between psychosocial factors and low back disorders (National Research Council, 2001). This review has shown evidence for a relationship between psychological work factors and future back pain. Specifically, evidence has been found for the relationship between low back disorders and job satisfaction, monotonous work, work relations, work demands, stress and perceived ability to work. In a systematic review by Hoogendoorn et al. (2008) strong evidence was found for low workplace social support and low job satisfaction as risk factors for LBP. There was insufficient evidence for high work pace, high qualitative demands, low job content, and low job control.

1.2 Whole-body vibration

Whole-body vibrations are mechanical energy oscillations transferred to the body. The magnitude of the vibration is determined by the extent of this motion, while the frequency is determined by the repetition rate of the cycles of oscillation. The term WBV is used to describe a situation when the whole-environment is undergoing motion, the body is supported on a surface that is vibrating and the effect of interest is not local to any point of contact between the body and the environment (Griffin, 1990).

WBV may be applied in three ways: 1) sitting on a vibrating seat; 2) standing on a vibrating surface; or 3) lying down on a vibration bed (Griffin, 1990). This project will specifically deal with WBV in the seated position. While seated, the feet are exposed to vibration from contact with the floor, the buttocks from contact with the seatpan, and the back from contact with the seat backrest. Health problems associated with WBV are dependent on a number of factors including vibration exposure magnitude, direction, frequency, and duration (Griffin, 1990; Mansfield, 2005).

1.2.1 Prevalence of low back pain and disorders associated with occupational WBV exposure

It is estimated that in Canada, the United States and some European countries, approximately 4 to 7% of employees are exposed to potentially harmful levels of WBV (Bovenzi, 1996). Wasserman et al. (1997) claim that in the United States, approximately 7 million workers are exposed to WBV.

Among various professional occupational drivers, complaints concerning the musculoskeletal system are most frequently reported in the neck, shoulders and lower back (Krause et al., 1997; Magnusson et al., 1996). Because of their driving, they are exposed to WBV. Through experimental studies, it has been found that resonance frequencies of the spinal column and other parts of the body lie between 1 and 10 Hz, which is the range of dominant frequencies found in occupational vehicles (European Committee for Normalisation, 1996). Studies have shown that occupational drivers report a relatively high prevalence of LBP in the range of 55-65% (Bovenzi et al., 1999; Bovenzi, Pinto, & Stacchine, 2002; Okunribido, Magnusson, & Pope, 2006).

Certain occupations such as mining (Eger et al., 2006; Kumar, 2004), construction (Cann, Salmoni, Vi, & Eger, 2003; Kittusamy & Buchholz, 2004), transportation (Cann, Salmoni, & Eger, 2004; Paddan & Griffin, 2002), agriculture

(Paddan & Griffin, 2002), and forestry (Rehn, Lundstrom, Nilsson, Liljeland, & Jarvholm, 2005; Sherwin, Owende, Kanali, Lyons, & Ward, 2004) are linked to increased injury rates due to WBV exposure. Workers in these occupations usually drive heavy equipment vehicles (HEVs) designed to execute specialized, heavy duty tasks. Drivers of HEVs are twice at risk of developing LBP compared to non-drivers (Waters, Genaidy, Viruet, & Makola, 2008).

1.2.2 Physiological and biomechanical effects related to whole-body vibration

The effects of WBV are due primarily to the fact that the human body is a complex biomechanical and physiological structure characterized by rigid and wobbling masses, all affected by sinusoidal motion (Cardinale & Pope, 2003). The physiological responses that have been observed resulting from WBV exposure are numerous. The following are examples that have been observed: cardiovascular (e.g., increased heart rate); respiratory (e.g. hyperventilation); endocrine and metabolic responses (e.g., changes in blood composition); motor processes (e.g., muscle reflexes); sensory system responses (e.g., alteration to the vestibular system); and skeletal (e.g., spinal/disc degeneration) (Griffin, 1990). Research has validated these effects and contributed information on other risk factors such as muscular fatigue, reduced balance, altered vestibular function (Seidel et al., 1980), impairments of the female reproductive system, and discomfort (Bongers et al., 1990; Seidel & Heide, 1986; Seidel, 1993; Wasserman et al., 1997). Current research emphasizes the particular importance on the effects on the spine (Thalheimer, 1996) resulting from WBV exposure.

Evidence arising from *in vitro* studies has shown that prolonged vibration exposure may cause spine pathology (and nociception) through mechanical damage, most notably to the vertebrae, vertebral endplates, intervertebral discs, and low back musculature (Wikström, Kjellberg, & Landström, 1994). However, the causal mechanism of the relationship between WBV and LBP and disorders is not known. One hypothesis of a causal mechanism is that WBV leads to instability of the spinal column. The stability of the spine is important to avoid intervertebral movements that could increase the risk of back problems. The following sections will discuss: 1) the importance of stability in spine biomechanics; 2) a model for spinal stability; 3) discuss existing biomechanical measures that indirectly measure the different components relating to spinal stability; and 4) the possible injury mechanisms after WBV exposure.

1.3 Spinal Stability

1.3.1 Definition of stability

Spinal stability is a term that is ambiguous in spinal biomechanics and the concept is interpreted in many different ways depending upon the context (Reeves, Narendra & Cholewicki, 2007; Einstein, 1999). Although efforts have been made to better define this concept, it seems that there is still no consensus. The following are some definitions that identify key elements. Mechanical stability is simply defined as the ability to withstand force or stress without significant alteration of the position of the joint (hypermobility of the spine is when there is a generalized increase in the range of intervertebral movements) or without material changes/damage to the tissue surrounding the joint. To enable the successful transmission of these forces, mechanical stability of the spinal system must be assured (Cholewicki & McGill, 1996). Stability may also be defined as the ability of the human system to return to its position of equilibrium following a small perturbation (Stokes, Gardner-Morse, Henry, & Badger, 2000). This is the theoretical definition of mechanical stability. From a clinical point of view, stability may also be seen as the ability of the spine under physiologic loads to limit patterns of displacement so as not to damage tissues surrounding the spinal column, not to irritate the spinal cord or nerve roots and to prevent incapacitating deformation or pain due to structural changes (White & Panjabi, 1978).

Unlike the definition of mechanical stability, the concept of clinical stability introduces the idea that the spine can have different levels of stability. The maintenance of spinal stability is extremely important in decreasing the chance of low back disorders. In the absence of muscles, an *in vitro* ligamentous lumbar spine is unstable at compressive loads of only 88 N (Crisco & Panjabi, 1992). However, the *in vivo* spine may endure values greater than 6000 N while participating in daily tasks (McGill & Norman, 1986), and up to 18000 N in competitive power lifters (Cholewicki, McGill, & Norman, 1991) due to the stability that is provided by the trunk musculature (Bergmark, 1989) and the neural system. To overcome the torque effects of an external load and to maintain stability, the spine is largely dependent on the trunk muscles. Muscle cocontraction aids in the prevention of spinal instability (Lavender, Tsuang, Hafezi, Andersson, & Hughes, 1992; Panjabi, Abumi, Duranceau, & Oxland, 1989). Antagonistic muscle contractile forces generate trunk moments to provide resistance to movement and increase stability while simultaneously loading the spine in compression (Cholewicki & McGill, 1996; Granata & Marras, 1995). Improved spinal stability may be achieved by cocontraction of the antagonistic muscles of the spine (Cholewicki, Panjabi, & Khachatryan, 1997; Granata & Marras, 2000).

1.3.2 Panjabi's model of the spinal stabilization system

Many factors have been suggested as possible contributors to lumbar segmental instability. Panjabi (1992) introduced a model of the spinal stabilizing system as the interaction of three conceptually different subsystems, but all functionally interdependent: 1) passive, 2) active, and 3) neural control (Figure 1.2).

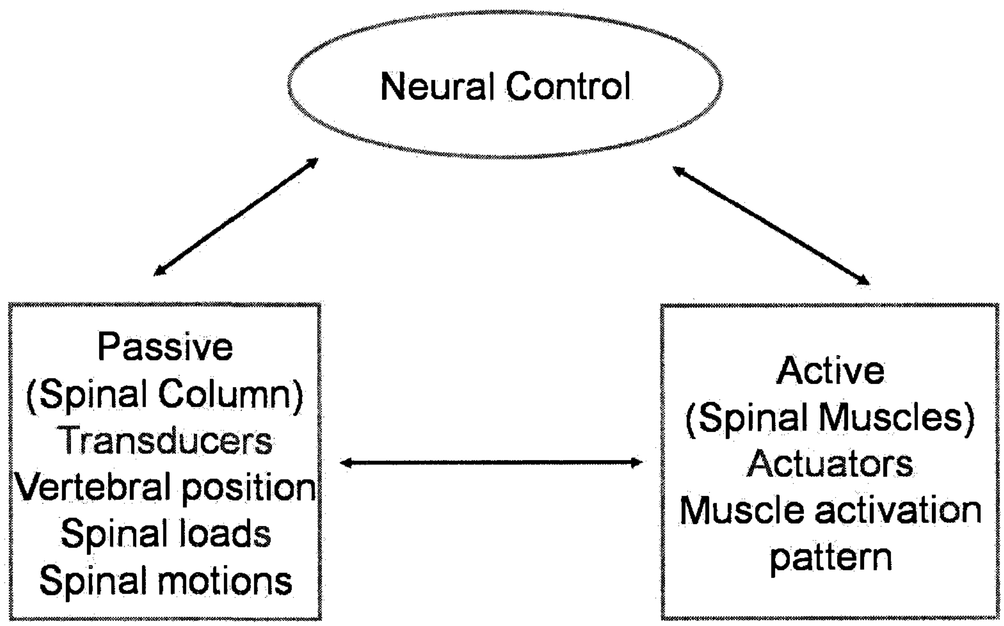

Figure 1.2: Interaction of the different subsystems in Panjabi's model of spinal stabilization

The passive subsystem is comprised of the vertebrae, facet articulations, spinal ligaments, intervertebral discs, and the passive mechanical properties of the muscles of the trunk. These elements of this subsystem do not provide much stability in the neutral position but rather act as transducers for measuring vertebral positions and motions. As the spine is moved out of the neutral region, tension develops in the passive structures, which is then measured by the central nervous system (CNS) to prevent end-range motions from occurring. The active subsystem is composed of spinal muscles and tendons surrounding the spinal column. The muscles generate the forces and the tendons measure the magnitude of the force generated in each muscle. Lastly, the neural control subsystem consists of neural components that receive information from various transducers and determines the specific requirements necessary for spinal stability.

The trunk muscles play an important role in spinal stability. An isolated spine cadaver (i.e., one that is stripped of all the muscles) buckles and therefore cannot support a load greater than 20 N. In addition, the lumbar portion of the spine has been shown to buckle under an axial load of 90 N (Crisco & Panjabi, 1992). These are only very small fractions of the loads that the spine must withstand to perform everyday normal activities. This demonstrates the importance of the trunk musculature in providing spinal stability. The muscles of the spine act to support structure and provide stiffness and stability to an otherwise unstable spinal column (Bergmark, 1989).

The spinal stability model proposed by Panjabi (1992) incorporates the idea of the "neutral zone", which is a region of laxity around the neutral resting position of a spinal segment. The neutral zone is based upon *in vitro* load displacement curves of a typical spinal motion segment. Instability can also be defined in terms of this neutral zone. Panjabi (1992) suggested that spinal instability is a likely result from dysfunction of either the spinal structures or trunk muscles or from reduced neural control. An increased neutral zone would therefore make the spine more instable. This neutral zone is shown to be larger with intersegmental injury and intervertebral disc degeneration (Kaigle, Holm, & Hansson, 1995; Mimura et al., 1994). An increase in trunk muscle activation has been shown to decrease the range of motion and the neutral zone of the trunk (Wilke, Wolf, Claes, Arand, & Wiesend, 1995).

An increase in stiffness would seem to be beneficial in providing stability but an increase in stiffness is the result of compression of the intervertebral disc (Janevic, Ashton-Miller, & Schultz, 1991). The increased compression forces could cause impingement of the bony tips of the zygopophyseal joints, thereby increasing friction forces between the articular processes of the lumbar vertebrae and then leading to an increase in resistance to motion (Janevic et al., 1991).

1.3.3 Measurements of spinal stability

Pope, Ogon and Okawa (1999) conducted a review of the literature of existing biomechanical measurements to determine spinal instability. These measurements include: 1) *In vitro experiments,* which evaluate the range of motion (ROM) of the functional spinal unit under different experimental settings; 2) *radiologic observations and measurement*, which are the most common methods to establish instability, but there is still controversy surrounding them; 3) *videofluoroscopy,* which has been used to measure three-dimensional motion in the sagittal plane during flexion-extension motion; 4) *palpation techniques* routinely used by manual manipulation practitioners, however the accuracy of this technique is less demonstrated than by other techniques; 5) *response to immobilization*, whereby spinal immobilization relieves pain, but the indications of the technique still need to be established and may be limited to a small subgroup of patients with disabling symptoms; and 6) *direct measurement of motion*, an invasive (though accurate) technique by directly placing pins in the spinous processes.

In the clinical setting, segmental spinal mobility is assessed by applying a posterior-anterior (PA) force to a single vertebral spinous process with the individual in the prone position. This technique correlates well with radiographic signs of instability. However, although it has good inter-tester reliability for identifying the least mobile segment, this assessment technique did not agree with sagittal-plane motion measured by dynamic magnetic resonance imaging and thus, questions the validity of the PA procedure for the assessment of intervertebral lumbar spine motion (Landel, Kulig, Fredericson, Li, & Powers, 2008).

Although several measurement techniques do exist, it remains difficult to measure spinal stability. These techniques either require exposure to ionizing radiation, require the subjects to remain as motionless as possible, are invasive, or have not been established as being reliable or valid.

This thesis uses Panjabi's lumbar stability hypothesis as a conceptual framework, implying that many injury pathways due to the different effects of WBV may be involved. Several indirect measures do exist to measure the different components of Panjabi's model. The following sections will discuss the measures to asses these components. Furthermore, they will discuss the effect of WBV and the possible injury mechanisms. The neuro-sensory functions such as back muscle reflex responses and postural balance may also be impaired by WBV, thus affecting the neural control subsystem. Whole-body vibration may have effects on the length of the spinal column, affecting the passive subsystem. Lastly, WBV may influence the muscles, representing the active subsystem of Panjabi's lumbar-stability hypothesis.

1.4 Assessment of the neural control subsystem

1.4.1 Reflex response

1.4.1.1 Sudden loading

Sudden loads can take many forms and are found both in daily and leisure activities, as well as in the workplace. Examples of sudden loading include slipping and tripping, unexpected slipping of an object being held in the hands or being lifted. Health care work has been associated with a high prevalence of low back disorders. Nurses handling patients (Burdorf & Sorock, 1991; Owen & Damron, 1984) and physical therapists (Molumphy, Unger, Jensen, & Lopopolo, 1985) have a higher probability of being exposed to sudden loading events due to the tasks that must be performed during patient handling.

The body is able to compensate for unexpected perturbations through several levels (mechanisms) of defense (Latash, 1998). First, there is the peripheral elasticity of the muscles, tendons and other tissues. During joint displacement, this elasticity provides instantaneous resistance against joint movement. Second, there is a stretch reflex, which also demonstrates visco-elastic properties and helps to dampen external perturbations, although at a certain reflex delay. The third level involves pre-programmed corrective reactions, or muscle activation patterns. This defensive mechanism has a longer delay (can occur as quickly as 100 ms following the perturbation), and is more powerful and more flexible than the first two mechanisms.

Sudden perturbations are hazardous by nature, however theoretically, there may be a lower risk of injury if a perturbation is expected rather than unexpected. If a person is able to anticipate the timing of a sudden load, the motor system may be able to coordinate and scale the muscle forces (Marras, Rangarajulu, & Lavendar, 1987) accordingly so that excessive force is not exerted and so that the system responds efficiently (Vera-Garcia, Elvira, Brown, & McGill, 2007). Likewise, anticipation of a loading allows the body to stabilize itself prior to the perturbation thereby minimizing trunk displacement (Marras et al., 1987) and muscle activity (Lavender & Marras, 1995). To maintain equilibrium of the trunk, the trunk muscles must be activated prior to the perturbation, given that the mechanical delay of the trunk muscles is more than 130 ms (van Dieën & de Looze, 1999). The preparations are in the form of anticipatory postural adjustements (APAs) (Aruin & Latash, 1995; Lavender et al., 1995) and cocontraction of the trunk muscles (Krajcarski, Potvin, & Chiang, 1999; Stokes et al., 2000; Thomas, Lavender, Corcos, & Andersson, 1998). APAs are prepared by the central nervous system CNS and are based on the information made available to the person (Shiratori & Latash, 2001). Cocontraction, though, is the simultaneous contraction of opposing muscle groups to maximize stability and to minimize the effects of a perturbation. It is not possible, though, to also anticipate perturbations as there are situations in our everyday lives where perturbations are also unexpected.

1.4.1.2 How is the reflex response measured in sudden loading applications?

Studies using the sudden loading paradigm have been carried out where the load is applied to the hand or the trunk to create an anterior perturbation of the upper body. In these studies, the subjects are either semi-seated/standing or standing with their pelvis either stabilized or not (Bull Andersen, Essendrop, & Schibye, 2004; Cresswell, Oddsson, & Thorstensson, 1994; Essendrop, Andersen, & Schibye, 2002; Gardner-Morse & Stokes, 2001; Herrmann, Madigan, Davidson, & Granata, 2006; Lavender et al., 1995; Skotte et al., 2004; Vera-Garcia et al., 2007). To control for as many confounding variables as possible, it is preferable to apply the load directly to the trunk and to stabilize the pelvis to restrict movement as much as possible to the lower body.

During the sudden loading test, measurement techniques such as electromyography (EMG) and trunk kinematics data have previously been collected (Bull Andersen et al., 2004; Herrmann et al., 2006; Krajcarski et al., 1999; Skotte et al., 2004; Vera-Garcia et al., 2007) to obtain simple measures associated to lumbar stability, these measures being: 1) EMG reflex latency; 2) EMG reflex amplitude; and 3) amplitude of the forward displacement of the trunk.

Documentation of the reliability of variables used in sudden loading studies is sparse, at best, in the literature. The reproducibility of a sudden loading test repeated over 10 trials was investigated (Skotte et al., 2004). Through an analysis of variance, the investigators found that the reaction time (latency) of the first trial was significantly longer than from trials 3-10. In this study, reliability coefficients were not calculated and visual determination of the EMG onset was used. Herrmann et al. (2006) measured spinal muscle reflexes using anterior-perturbations that were applied, while subjects were standing quietly, using a pendulum suspended by the ceiling. Three trials before and after a fatiguing task were performed. The reported intra-class correlation coefficient (ICC) for reflex delay and amplitude were 0.41 and 0.61, respectively. Unfortunately, we do not know how many trials would be necessary to increase these ICC values to a more acceptable level of reliability. Thus, this is an important step that must be taken before this method can be used to come to conclusions regarding the reflex response of back muscles following WBV.

1.4.1.3 The effect of whole-body vibration on the muscle reflex response

Very few studies have investigated the reflex responses due to a sudden load after WBV exposure. In the one known study of this nature, Wilder et al. (1996) established that the response latency and the magnitude of the response of the erector spinae muscles increased after exposure to vertical vibration (5 Hz frequency, 0.223 ms^{-2}

rms acceleration) for 40 minutes. They attributed the increases to muscular fatigue but this hypothesis was not substantiated. However, when they allowed the subject to walk for 5 minutes around the lab after the vibration exposure, the reflex latency and peak EMG muscle activity had a tendency to decrease, indicating that this time was enough time for recovery. More recently, Li, Lamis and Wilson (2008) found that the reflex response increased from an average of 205 ms to 228 ms after 20 min of vertical vibration (5 Hz frequency, 0.315 ms^{-2} rms acceleration). No comparison to a sitting only condition was made. These responses, however, are longer than 150 ms, meaning that they are no longer reflexive but rather voluntary responses (Schmidt, 1991).

Studying the acute effects of WBV on the back muscle reflex response is a research area that should be further pursued considering the documented risk of low back problems due to exposure to WBV and sudden loading events. One mechanism of injury that could explain the link between WBV and low back disorders is proposed in the next section.

1.4.1.4 Possible mechanism of injury

Along with the intrinsic stiffness of the active muscles (Bergmark, 1989), the reflex response (Solomonow, Zhou, Harras, Lu, & Baratta, 1998) is the primary mechanism for neuromuscular control of spinal stability. These reflex reponses are needed during sudden undesired vertebral motion (localized spinal buckling). The reflex response is a means of feedback control contributing to spinal stability (Cholewicki & McGill, 1996; Granata, Wilson, & Padua, 2002). WBV may compromise the reflex response (reducing the reflex gain or increasing the latency) thus reducing spinal stability. This reflex response could be important during the WBV exposure itself (i.e., there may be a jarring motion of the trunk during exposure), but if the effect of WBV continues after exposure, the trunk is vulnerable during other occupational tasks such as MMH.

1.4.2 Postural control and balance

Postural control when maintaining upright stance can be defined as the body's ability to maintain the body's centre of mass (COM) within the base of support by counteracting gravitational or external forces as well as internal forces produced by voluntary movements (Massion, 1992; Winter, Patla, & Frank, 1990). The postural control system is responsible for processing the sensory input signals from the vestibular, visual, and somatosensory systems and the central nervous system (CNS) generates the appropriate muscular control signal required to maintain balance and upright stance.

Recent models have suggested that postural control can be controlled at a lower level simply by manipulating ankle stiffness (Winter, 1995b; Winter, Patla, Prince, Ishac, & Gielo-Perczak, 1998). This model has been named the "Inverted Pendulum Model" (Winter, 1995a). According to Winter et al. (1998), the CNS controls the COM of the body by setting the appropriate muscle tone. This determines the joint stiffness needed to maintain upright stance in a particular situation. If the body is modelled as an inverted pendulum, the centre of pressure (COP) under the feet can be modified in order to maintain the COM within the base of support. Therefore, COM is the controlled variable and COP is the controlling variable (Winter et al., 1998).

1.4.2.2 Measuring balance

Postural steadiness is often characterized by postural (or body) sway (Prieto & Myklebust, 1993; Prieto, Myklebust, Hoffman, Lovett, & Myklebust, 1996), a kinematic term often estimated from center-of-pressure (COP) measures derived from force plate data (Winter et al., 1998). The characteristics of the movement of the COP, defined as the point of application of the ground reaction forces under the feet (Winter, 1995a), have been used to infer neurological and biomechanical mechanisms of postural control among different populations (De Haart, Geurts, Huidekoper, Fasotti, & van Limbeek, 2004; Goldie, Evans, & Bach, 1994; Lacour et al., 1997; Lord & Sturnieks, 2005; Melzer, Benjuya, & Kaplanski, 2004; Mientjes & Frank, 1999; Trenkwalder et al., 1995). The study of the path of the COP from a single platform in a laboratory setting is a common outcome measure in research in quiet standing (Winter, 1995b).

Several summary based variables on the COP have been widely used. Goldie, Bach and Evans (1989) suggested that these outcome parameters may measure different aspects of posture. A few examples will be given in the following sentences. The sway path is a measure of the total distance of the COP displacement, however it does not provide insight into how posture is controlled in the anterior-posterior (AP) versus medio-lateral (ML) direction. The maximum range of the COP displacement (in the AP and ML directions) focus only on the large exertion of the COP and does not offer a precise representation of the average control. Measures of velocity offer information about the tightness of the control. The root-mean-square (RMS) distance gives an idea of the sway amplitude. Winter (1995b) introduced a new variable, the COP-COM difference, which is proportional to the horizontal acceleration of the COM during quiet standing.

Like many biological measurements, the COP has an intrinsic variability that affects the reliability and validity of postural control outcomes. The test-retest reliability of various COP-based summary measures has been studied by computing intra-class correlation coefficients (ICC) among pairs of scores obtained over repeated measurements (Carpenter, Frank, Winter, & Peysar, 2001; Lafond, Corriveau, Hébert, & Prince, 2004; Mientjes & Frank, 1999).

Carpenter et al. (2001) investigated the optimal sampling duration time between testing days. Using young, healthy subjects, 120-second trials (n = 3) were performed with eyes open. The trials were collapsed into 15 s, 30 s, 60 s and 120 s trials and then the AP and ML values of the RMS, mean power frequency (MPF) and mean position (MPOS) were computed. MPOS was the most reliable, with an ICC of 0.89 and 0.84 for AP and ML, respectively. The least reliable was MPF, with an ICC of 0.45 and 0.31 From this study, they recommended a sampling frequency of at least 60 seconds. Therefore, if this is the case, then this would put into question the results obtained from Goldie et al. (1989) and Le Clair and Riach (1996) who found good test-retest reliability scores with sampling durations of 15 s and 20-30 s, respectively.

Lafond et al. (2004) aimed to look at the intra-session reliability of six common COP summary variables in healthy elderly subjects during quiet standing with their eyes open. They found that excellent reliability was achieved with COP velocity (ICC 0.77-0.90) by performing only one trial of quiet standing for 60 s. They also determined that at least 10 trials needed to be averaged to have an ICC of at least 0.90 for the following variables: sway area, COP range (AP), MPF and MedPF. Corriveau, Hébert, Prince and Raîche (2000) estimated the intra-session reliability of the COP-COM difference and concluded that 4 trials should be averaged to obtain a reliable measure.

Mientjes and Frank (1999) performed a reliability analysis on only a few COP measures (RMS, MEAN, MPF) in their study of balance between healthy people and low back pain patients under seven different conditions. The estimated ICCs ranged between –0.22 and 0.89, depending on the measure and condition.

It appears that there is no widespread consensus of the reliability of the COP-based summary measures. However, this is largely due to the differences in techniques, methods, and interpretation of the analysis. It would also be interesting to document the reliability of substantial set COP summary measures.

1.4.2.3 Effect of whole-body vibration on balance

Studies examining the effects of WBV on postural balance are also limited (Martin, Gauthier, Roll, Hugon, & Harlay, 1980; McKay, 1972; Seidel et al., 1980). Exposure to WBV has resulted in increases (though not always significant) in postural sway at fixed sinusoidal frequencies ranging from 12.5 Hz – 18 Hz (Martin et al., 1980;

McKay, 1972). The external validity (i.e., in relation to the workplace conditions experienced by drivers) of these studies could be questioned with this use of only pure sinusoidal WBV. This issue was addressed in a study that exposed subjects for 40 min to vertical WBV with a frequency spectrum resembling a shuttle car operation (Cornelius, Redfern, & Steiner, 1994). They found no differences before and after WBV on any of the balance measures. However, only one trial was performed, thus limiting the reliability and sensitivity of this measure.

1.4.2.4 Possible injury mechanism(s)

One mechanism may be if the vestibular system, one of the three major sensory systems involved in balance, is compromised in a negative manner, then there may be a higher risk for loss of balance. Vibration in the vertical direction was previously shown to affect the vestibular system (Suvorov et al., 1989). A second mechanism is explained by a change in the activation of the secondary endings of the muscle spindles. It has been demonstrated that vibration is an effective mode of activating muscle spindles (Burke, Hagbarth, Lofstedt, & Wallin, 1976; Goodwin, McCloskey, & Matthews, 1972) and the effector site of this vibration stimulation is located in the secondary endings of the muscle spindles which is considered to regulate posture (Eklund, 1973). In seated whole body vibration, vibration may be transmitted through the surface of the vibrating surface and any points of the body that are in contact with the surface (Griffin, 1990). Thus, in the seated position, it could be through the feet, buttocks, back of the thighs, and even the back (if there is a backrest on the seat) of the individual. This may cause indirect vibration to the leg muscles and hence may affect standing posture. Yagi, Yajima, Sakuma and Aijara (2000) observed significant sagittal body sway when the triceps surae, tibialis anterior, biceps femoris muscles were vibrated.

1.5 Assessment of the passive subsystem

When compressive loading is sustained there is a gradual deformation of collagenous structures over time (Bogduk, 1997). This phenomenon, called creep, is a normal diurnal occurrence in mammals. One mechanism to explain this is because water is slowly expelled from the loaded tissue. This can be reversed: when loading is reduced, the expelled fluid simply flows back in again, rapidly at first, but then slowing down later (Adams, Bogduk, Burton, & Dolan, 2002).

Human stature undergoes diurnal changes (Botsford, Esses, & Ogilvie-Harris, 1994; Reilly, Tyrrell, & Troup, 1984) and an individual's height is usually less at the end compared to the beginning of the day. It has been demonstrated that overall height loss throughout the day varies up to 15-20 mm (Krag, Cohen, Haugh, & Pope, 1990; Tyrrell, Reilly, & Troup, 1985). The majority of stature loss has been attributed to alterations in intervertebral disc height (Adams & Hutton, 1983; Foreman & Troup, 1987). This decrease in the disc height and consequently decrease in overall stature has often been termed as "spinal shrinkage." Spinal shrinkage due to spinal compression is thought to be caused by a combination of fluid loss from the motion segment and viscoelastic deformation (van Dieën & Toussaint, 1993; van Dieën, Creemers, Draisma,

& Toussaint, 1994). It is estimated that two-thirds of height loss is attributed to fluid loss (Adams et al., 1983).

1.5.1 Measuring change in stature

Direct measurements of spinal loading through *in vivo* studies are normally avoided because of concerns of introducing a transducer into the disc. Just over 20 years ago, Eklund and Corlett (1984) popularized the use of stadiometry (using an apparatus called a stadiometer) as a non-invasive and inexpensive technique to measure height-change and to evaluate cumulative loading effects on the spine for various work tasks and postures. Recently, van Deursen, van Deursen, Snijders and Wilke (2005)found a good correlation between spinal shrinkage using a stadiometer and the intradiscal pressure (IDP) method. They concluded that use of spinal shrinkage measurement appears to be a good alternative for IDP measurements in static situations.

This technique has been widely used in research to reflect spinal deformation under loaded and unloaded conditions (Althoff, Brinckmann, Frobin, Sandover, & Burton, 1992; Leivseth & Drerup, 1997; McGill, van Wijk, Axler, & Gletsu, 1996; Tyrrell et al., 1985; van Dieën & Toussaint, 1993). This method has widely been used to investigate the effects of repetitive symmetrical (Stålhammar, Leskinen, Rautanen, & Troup, 1992; Tyrrell et al., 1985; van Dieën et al., 1994) and asymmetrical (Au, Cook, & McGill, 2001) lifting, axial compression (Althoff et al., 1992; Kanlayanaphotporn, Trott, Williams, & Fulton, 2001; Kanlayanaphotporn, Williams, Fulton, & Trott, 2002; Tyrrell et al., 1985) and whole-body vibration (see following section).

On average, it has been suggested that a standard deviation (SD) of 0.5 mm between repeated successive measures be used as an acceptable level of repeatability for measuring changes in stature (Rodacki, Fowler, Rodacki, & Birch, 2001). Sullivan and McGill (1990) claimed to have reliable measures with the subject seated with a reported SD of 1.4 mm. However, these measures were taken using a meter stick and the spinal curvature was not controlled for, thus could potentially explain the large SD compared to the accepted 0.5 mm. Others have reported SDs of 0.4 – 0.9 mm (Eklund & Corlett, 1984; Klingenstierna & Pope, 1987; Leivseth & Drerup, 1997) by measuring in the standing position.

However, very few studies have focused on the repeatability of these measures. A study by Rodacki et al. (2001) examined the repeatability of measurement and the number of trials necessary to obtain an acceptable level of reproducibility in measurements of spinal length in both standing and sitting postures. They found that repeatability was achieved more quickly in the standing posture than the sitting posture. That is to say that it took less series (two versus 3) of 10 measurements to achieve a mean SD of at most 0.5 mm.

Kanlayanaphotporn et al. (2002) computed three reliability coefficients to reflect the reliability of the creep response in asymptomatic and low-back pain subjects. They computed the intraclass-correlation coefficient (ICC) to show the level of consistency and agreement of responses among subjects, the standard error of measurement (SEM) to reflect the random variability of a single individual's values on repeated testing, and

SD. The ICC ranged from 0.56-0.91 and 0.07-0.89 for the asymptomatic and LBP subjects, respectively. This range in values was explained by the fact that the range of spinal creep response for each measurement time of each day became wider, with larger SDs, as time increased. Accordingly, differences among subjects increased and ICC values increased with time. The SEM values ranged from 1.02-1.73 mm and 1.00-1.98 mm and the SD ranged from 0.86-1.49 mm and 0.83-1.77 mm for the asymptomatic and LBP subjects, respectively.

The different measurement protocols used in stadiometry studies accounts, at least in part, to the differences in the results obtained. Stothart and McGill (2000) found less variability in their measures by leaving the subject in the stadiometer during repeated measurements versus the “in-out” method (i.e., having the subject step off the stadiometer between each measure). However, this could also introduce systematic error in the measurement technique. However, even in similar loading conditions, there is still variability in subject response, particularly between days, and thus this is a limitation to using this type of measurement. McGill et al. (1996) suggest different factors and relationships that may influence the inter- and intra-subject variability in spinal shrinkage. Some of the factors include age, gender, disc area, existing injury, loading history, anthropometrics and anatomical variables (i.e., height, weight, strength) and disc mechanics (i.e., fluid content and fibre condition).

Therefore, there is still more research needed to improve this measurement technique to reduce variation in the measurements. Better controlling for and identifying different sources of measurement error could improve the reliability of this measure.

1.5.2 Effects of whole-body vibration on stature

Sullivan and McGill (1990) found that there was a decrease of 9 mm in spinal height after 30 minutes of vibration at a frequency of 5 Hz, compared to a 1 mm shrinkage from the control. They also found that these subjects were taller at the end of the day compared to a control group. They propose two possible mechanisms to explain why the spine lengthens, when over the course of the day it normally shrinks. The first is due to the viscoelastic properties of the joint ligaments and annulus. The tissues may stretch during vibration and remain elongated the rest of the day until normal resting length is regained. A second is that there is an inflammatory response. After mechanical injury, the blood vessels dilate increasing the loss of fluid into the surrounding tissue, accomplished by increased permeability of the vessel walls to protein. Thus, swelling occurs due to the additional fluids and an increased osmotic pressure in the nucleus.

Magnusson, Almqvist, Broman, Pope, and Hansson (1992) exposed subjects to a vibration frequency of 5 Hz and 0.1 *g* RMS acceleration. Both vibration and no vibration conditions were performed on the same day. Six, five-minute exposures of alternating vibration and no-vibration were performed. Measurements were taken after each set of five minutes. They found that there was significant height loss due to vibration (5.94 mm) compared to no vibration (4.52 mm). After controlling for the posture, they found that postural change was responsible for approximately 50 % of the total height loss, thus emphasizing the importance of controlling for posture as it influences the accuracy of the measurements and the interpretation of the results. Similarly, subjects exposed to whole body vertical vibration showed a height loss directly after vibration with a return to the height expected for the time of day within a few hours (Klingenstierna et al., 1987). These subjects were exposed to a vibration frequency of 5 Hz and acceleration of 2 m/s^2 for a period of 30 minutes. Magnusson et al. (1992) offer the explanation that increased disc loading due to the increased transmissibility at the resonant frequency is the most likely explanation for the measured creep with vibration.

Contradictory results, however, have also been found. Althoff et al. (1992) found, however, that sitting on a chair without a backrest under vertical vibration resulted in an increase in stature compared to standing. However, they showed no difference in stature change due to sitting alone or sitting with vibration exposure. Bonney and Corlett (2003) found that exposure to 60 minutes of sitting with no vibration and vibration at a frequency of 8 Hz caused spinal shrinkage of 1.19 mm and 0.03 mm, respectively. However, at vibration frequencies of 4 Hz and 6 Hz, an increase in height of 1.76 mm and 0.05 mm, respectively, were observed. The vibration was both horizontal and vertical and thus, bi-directional vibration exposure could result in unloading of the spine at frequencies close to the natural frequency of the human body.

1.5.3 Possible injury mechanism(s)

It has been demonstrated that changes in body height may be used as a measure of vertebral disc flattening, reflecting spinal loading, which when excessive, may lead to low back injury. One hypothesis is that whole-body vibration would increase mechanical loading and forcing the fluid out of the interevertebral discs and causes intervertebral height loss. The rest of the height loss is due to viscoelastic deformation that can occur in the vertebral end plates (Brinckmann, Frobin, Heirholzer, & Horst, 1983) and sideways bulging of the annulus fibrosus (Reuber, Schultz, Denis, & Spencer, 1982). Eventually, intervertebral ligaments and the posterior fibres of the annulus become more slack and are less able to resist sudden flexion movements. This could possibly result in increased joint movement, and thereby increasing risk of injury.

1.6 Assessment of the active subsystem

1.6.1 Muscular fatigue

Neuromuscular fatigue has been defined as "a general concept intended to denote an acute impairment of performance that includes both an increase in the perceived effort necessary to exert a desired force and an eventual inability to produce this force" (Enoka & Stuart, 1992). Muscle fatigue is a reduction in force that a muscle can generate or when a muscle can no longer maintain the required force due to exercise. Fatigue is task-dependent and thus the task designates the underlying mechanism(s) and also the site(s) of fatigue.

1.6.2 Measuring muscular fatigue

Several methods exist for assessing neuromuscular fatigue. They will be summarized here, however a full description may be found in Vøllestad (1997).

1.6.2.1 Direct Measures

Maximal voluntary force contraction: This method is often used in humans and is considered the "gold standard". Reliable assessment is highly dependent on the force generating capacity. However, the force generated voluntarily can be limited by the lack of motivation by an individual.

Power output: The ability to generate power may be as or even more important than the ability to generate force. In fatigue studies, changes in power output are examined from the temporal change in power of each contraction through a short maximal effort. However once again, this is dependent on the level of motivation of the individual.

Electrostimulation (titanic force): Maximal force or power is examined by electrical stimulation of the motoneurones or the muscle itself. This method abolishes any limitations in the central nervous system and is a direct measure of peripheral fatigue. That is to stay, a direct measure of the capacity of the muscle in question may be obtained.

Low frequency fatigue: Many fatigue studies use twitch force as an estimate of the loss of force generating capacity. A disproportionate fall in twitch force (i.e., needing hours or even days to recover completely) is called low-frequency fatigue and is reported during high-intensity exercise as well as during submaximal repetitive contractions.

1.6.2.2 Indirect Measures

Twitch interpolation: This technique is based on assessing the twitch contraction elicited by either a single or a double electrical stimulus delivered to the muscle or nerve during a contraction. The force increment in response to the stimulus reflects the force reserve. This method provides evidence for central fatigue.

Endurance time: This approach presumes that there is an association between the decline in maximal force generating capacity and the time to exhaustion. However, it has been shown that the relationship between these two parameters varies. Thus, there are different mechanisms behind the development of fatigue and exhaustion. This approach is also dependent on the level of motivation of an individual.

Electromyography: Surface electromyography (EMG) is commonly used for the examination of muscular reactions and is one indirect objective and non-invasive measure of muscular fatigue. Surface electrodes pick up the electrical activity of superficial muscles and the amplitude and power spectrum of the signal may be determined. The amplitude reflects the number and size of action potentials in the muscle over a given period (Basmajian & De Luca, 1985). Fatigue induces many changes in the action potentials and in the EMG of contracting muscles. Briefly, fatigue may manifest itself as increase in the EMG amplitude [time domain] (Arendt-Nielsen &

Mills, 1988) and/or a shift to lower frequencies [frequency domain] (i.e., decrease in the mean power frequency: MPF) of the power spectrum (Dolan, Mannion, & Adams, 1995).

1.6.3 The effect of whole-body vibration on muscle fatigue

The muscles, which are part of the active subsystem, are influenced by vibration. The premise is that exposure to WBV may result in muscle fatigue due to acute reflex activation of the primary muscle spindle fibres. Mechanical vibration (namely between 30 and 120 Hz) directly applied to a muscle belly or tendon elicits the tonic vibration reflex (TVR) (Desmedt, 1983; Vermeersch, Vermeersch, & Vermeersch, 1986), a neuromuscular response caused by excitation of muscle spindles leading to enhanced muscle activity. The TVR has been suggested to occur in the back muscles at frequencies between 1 and 5 Hz (Seidel, 1988), however, the evidence to support this phenomenon happening at lower frequencies is still not conclusive. This increased muscle activity is necessary to dampen the vibratory waves (Wakeling & Nigg, 2001), though could lead to muscle fatigue.

Few laboratory studies using surface EMG have demonstrated back muscle fatigue after exposure to seated WBV (Hansson, Magnusson, & Broman, 1991; Wilder, Magnusson, Fenwick, & Pope, 1994). The shortcomings of these studies are the short exposure duration (i.e., less than 10 minutes) and different conditions performed on the same day (Wilder et al., 1994), which makes it difficult to separate the effects of a single condition. Hansson et al. (1991) found that the mean frequency of the EMG signals obtained from the erector spinae muscles decreased over time and the root-mean-square (rms) values increased. However, to ensure that muscular activity was present, they had the subjects leaning forward in a bent position and carrying extra weight on the front of their chest while they were sitting. This is not representative of realistic working conditions.

Contrary to the above findings, a study on helicopter pilots in flight (de Oliveira & Nadal, 2004) found no back muscle fatigue as revealed by the slope of the linear regression of the median frequency. EMG of the left and right erector spinae muscles were recorded for 2 hours. This might be explained by the low mechanical exposure induced by the vibration. For 88% of the pilots, for 50% of the time, their back muscles were below 5% of their maximal voluntary contraction (MVC). For 90% of the time, the EMG activity was below 14% MVC. Jonsson (1978)defined three levels of muscle load: static, dynamic, and peak, based on the Amplitude Probability Distribution Function (APDF). The APDF curve is the distribution of the levels of muscle contraction during the observation period. When graphed, it can be used to identify the percentage of time that muscle activity is less than a given proportion of the person's maximal voluntary contraction (MVC). Jonsson proposed threshold limits in cases of continuous work: the 10^{th} percentile (static) should not exceed 2-5% MVC; the 50^{th} percentile (dynamic, also referred to as the "mean") should not exceed 10-14% MVC; and the 90^{th} percentile (peak) should not exceed 50-70% MVC.

1.6.4 The effect of fatigue on spinal stability and possible mechanisms of injury

As defined in section 1.6.1, muscle fatigue is the inability to generate the required level of force. Therefore, fatigue muscles will decrease their force output. Another effect could be a decrease in the muscle reflex response (reflex latencies). Several studies have investigated the effect of fatigue on reflex response, however, the results are still inconclusive. Some studies have found no effect (Granata, Slota, & Wilson, 2004; Herrmann et al., 2006), while others found shorter reflex latencies (Magnusson et al., 1996; Mawston, McNair, & Boocock, 2007). Wilder et al. (1996) found that when fatigue of the erector spinae muscles was induced by WBV, the reflex response of the muscles increased.

Balance may be affected by fatigue because of proprioceptive inhibition, or, in cases of severe fatigue, because the muscles are so fatigued that they are unable to generate enough force to maintain balance (Johnston, Howard, Cawley, & Losse, 1998). However, the former reason seems more probable. Johnston et al. (1998) found that following a closed kinetic chain antagonistic exercise (similar to a stair stepper), fatigue significantly decreased balance during three static balancing tests. Nardone, Tarantola, Giordano and Schieppati (1997) found that sway area and sway path significantly increased following a treadmill fatiguing session.

Davidson, Madigan and Nussbaum (2004) investigated the effect of lumbar extensor fatigue, fatigue rate, and fatigue recovery on quiet standing. They found an increase of up to 58% in the time-domain measures but no changes in the frequency-domain measures. The rate of fatigue did not affect the magnitude of the postural sway increases, nor was the rate of balance recovery following fatigue affected.

Fatigue may affect the central nervous system and have indirect effects by altering muscle coordination. It is well documented that fatigue reduces neuromuscular control of trunk movement by increasing the variability in movement patterns (Ng, Parnianpour, Richardson, & Kippers, 2003; Parnianpour, Nordin, Khanovitz, & Frankel, 1988) and by increasing the variability in muscle activation patterns (Ng et al., 2003; van Dieën, Cholewicki, & Radebold, 2003). Therefore, as muscle fatigue occurs, the neural control systems places more attention on the ability to continue the task performance, rather than the quality of the performance and the stability requirements of the spine.

Spinal stability is primarily controlled by muscle recruitment, active muscle stiffness, and reflex response. Neuromuscular fatigue is one factor that may reduce neuromuscular control of the spine, and thus represent a threat to its mechanical stability.

CHAPTER 2

THESIS OUTLINE

To answer the specific research questions and address the objectives, three separate studies were conducted in this dissertation. This thesis consists of a collection of three peer-reviewed (two accepted and one at its first revision) journal articles (Chapters 3, 4, and 5) and a presentation of results (Chapter 6) from the three studies. The relationship of the three studies conducted in this dissertation is presented in Figure 2.1 and briefly described in the paragraphs that follow.

The introduction includes a brief background of the issues that laid the foundation for this thesis and the general purpose. To facilitate interpretation and implementation of these findings, the introduction includes the rational and underlying issues for examining the biomechanical responses to seated WBV, and the literature review discusses the issues associated with the work in this thesis. The dissertation concludes with a general discussion and summary (Chapter 7) that provide an overview and integration of the main findings of each study. The major contribution of each study is also highlighted in the summary.

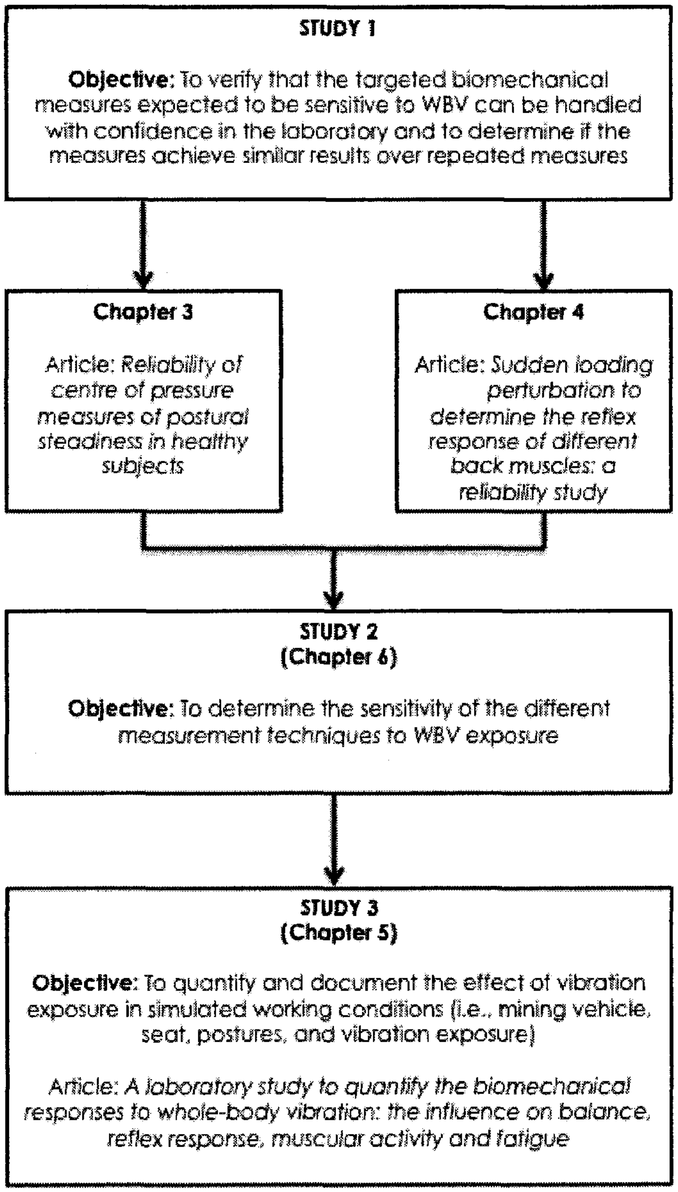

Figure 2.1: Overview of the relationship between each study in this dissertation

Three experimental studies were conducted. These studies, as well as the specific role of each chapter will be described. The literature review (Chapter 1) provides an overview of the major issues with this dissertation and pertinent findings. This chapter

(Chapter 2) provides an overview of the relationship of the three experimental studies. The first study focused on the reliability of various biomechanical measures that were chosen to study the effects of WBV. The following chapters (Chapters 3 and 4) were built on the findings reported in the first study.

The second study investigated the sensitivity of the biomechanical measures (to seated vertical WBV). The effects of the measures were examined after a period of 60 minutes sitting only and 60 minutes of WBV. The vibration magnitude was one in which the dominant frequency of the signal was close to the human whole-body fundamental resonant frequency. Participants were seated on a rigid seat with no backrest, mounted on a whole-body vibration vehicular simulator. The basic idea was to test the effect of WBV under "extreme" conditions (i.e., magnitude of vibration exposure near the resonant frequency, rigid seat), while still below the limits of exposure set by the International Organization for Standardization, on the sensitivity of the different measures. The results of this study are presented in Chapter 6. It was decided to first publish the results from Study 3 because the WBV exposure corresponded to field conditions in the mining industry. This explains why Study 2 results are presented at the end. A discussion of the results is integrated in the general discussion (Chapter 7).

The third study used many of the same biomechanical measures as in the second study to quantify the effects of WBV, but under more realistic occupational exposure conditions. Results of this study are presented in Chapter 5. Biomechanical responses were tested before and after 60 minutes of sitting, with and without vertical whole-body vibration (WBV). To increase the face validity of the measures, the vibration acceleration magnitudes to which the participants were exposed were those of a large mining load haul dump (LHD). The signals that were simulated in the lab were taken from field measurements. The method by which the spectral class characteristics were determined is explained in Appendix C. Similarly, postures adopted by the participants while sitting were selected based on postures adopted by LHD operators.

CHAPTER 3

RELIABILITY OF CENTRE OF PRESSURE MEASURES OF POSTURAL STEADINESS IN HEALTHY SUBJECTS

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3.1 Abstract

This study aimed to 1) estimate the reliability of 36 center of pressure (COP) summary measures in healthy subjects and 2) identify the main sources of variability in order to estimate the most appropriate measurement strategies to improve reliability. Twelve healthy males performed, on two separate days, eight one-minute trials of quiet standing on a force platform in two conditions [eyes-open (EO) and eyes-closed (EC)]. The generalizability theory was used as a framework to estimate the magnitude of the different variance components (Subject, Trial, Day, and all interactions) and the reliability of the measures corresponding to various simulations of measurement strategies. Reliability of the COP summary measures was poor to moderate. Intra-class correlation coefficients were generally higher with EO (mean: 0.46, range: 0.03 - 0.76) than with EC (mean: 0.41, range 0.02 – 0.72) across all summary measures. The majority of the variance was attributed to Subject (2% - 76%), Subject x Day (0% - 24%) and Subject x Day x Trial (16% - 79%) variance components depending on the summary measure and condition. The reliability could be improved more efficiently by averaging measurements between-days than by increasing the number of trials during one day. For the majority of the summary measures, acceptable reliability can be achieved when at least 7 or more trials are averaged during the same testing day.

Keywords: reliability, centre of pressure, measurement strategies, postural steadiness

3.2 Introduction

Postural steadiness is often characterized by postural (or body) sway [1,2], a kinematic term often estimated from center-of-pressure (COP) measures derived from force plate data [3]. The characteristics of the movement of the COP, defined as the point of application of the ground reaction forces under the feet [4], have been used to infer neurological and biomechanical mechanisms of postural control among different populations [5-11]. The study of the path of the COP from a single platform in a laboratory setting is a common outcome measure in research in quiet standing [12]. However, like many biological measurements, the COP has an intrinsic variability affecting the reliability and validity of postural control outcomes. Therefore, the reliability of COP measures should first be established before they are used to either monitor if a patient's balance improves over the course of a clinical intervention and/or to evaluate standing balance for the diagnosis of different pathological populations.

Studies reporting the reliability of the traditional COP variables (e.g., RMS, mean COP velocity, MPF, MedPF, range of sway, fractal dimensions) [10,13-19] differ according to the assessed COP variables but more importantly, according to the sources of variability considered. Corriveau et al. (2000) give a comprehensive explanation of three different types of variability that can contribute to measurement error and thus, affecting different types of reliability: intrasession (within the same testing day), intersession (between testing days), and interrater (between raters or experimenters). Reducing these sources of variability, would improve reproducibility and responsiveness, and thereby reliability [20]. Interrater reliability is unlikely to be of concern for the measurement of COP due to the simplicity of the apparatus, task and instructions. In studies that have investigated intrasession reliability, recommendations made as to the number of trials needed to be averaged during a single testing session, as well as the trial length to obtain acceptable reliability has differed [13,21] depending on the measure. To the authors' knowledge, only one study, which recommended an optimal trial length, on standing balance has evaluated intersession reliability (i.e., test-retest, one rater) [17].

The most common index used to report the reliability is the intra-class correlation coefficients (ICC). The ICC, or the ratio of the variance between subjects to total variance, is often calculated within the framework of the *classical test theory*. However, this theory does not allow the partitioning of the other sources of variance influencing measurement error (i.e., systematic and random). To address this limitation, the generalizability theory (G theory) was developed [22]. The G theory allows an investigator to estimate the magnitude of the different sources of error contributing to the measurement error and then design measurement strategies to try to reduce this error and improve reliability. Only one study [21] has used this theory to estimate the reliability of a limited number of COP measures (n=4) of quiet standing.

Using the G theory, this study will 1) determine the reliability of various COP summary measures [2] obtained from healthy subjects and 2) identify the main sources of variability in order to estimate the most appropriate measurement strategies to improve reliability. This will be carried out for eyes open and eyes closed to quantify the effect of vision.

3.3 Methods

3.3.1 Participants

Twelve healthy males, recruited from a student population, participated (mean age, height, and weight: 26.9 ± 4.7 years, 1.75 ± 0.07 m, and 74.9 ± 13.1 kg, respectively). They reported to be free of neurological illness, musculoskeletal disorders, degenerative conditions or any disease that would interfere with their normal balance. Participants read and signed an informed consent form approved by the university Research Ethics Committee.

3.3.2 Equipment and Procedure

Ground reaction forces were recorded with at a force platform (BP900900, Advanced Mechanical Technology, Inc., Watertown, MA) at a sampling frequency of 100 Hz and then converted to a digital signal via a 16-bit A/D converter. Participants stood barefoot on the surface of the force plate with both feet parallel on both sides of a 5.1 cm T-shaped separator placed on the surface of the force plate. This separator was always placed at the same position on the force plate and then removed once the participant's feet were positioned. The participant was then instructed to stand quietly with arms hanging to their sides and looking forward.

One data collection session required that the participant perform eight (n=4 eyes-open [EO], n=4 eyes-closed [EC]) 60-s quiet standing trials. After each trial, the participant stepped off the force plate then immediately stepped back onto it and positioned himself for the next trial. During the EO condition, the participant's eyes were focused on a stationary target (at approximately eye-level) located 2 m from the center of the force plate. The conditions were presented in a counterbalanced design. To assess the intersession reliability, each participant returned to the laboratory, no later than one week after the first visit, and performed the same procedure.

3.3.3 Data analysis and computation of COP-based summary measures

The COP was computed using the force plate outputs (forces, moments) using an in-house C++ program. Using MATLAB (The Mathworks, Natick, MA), the COP time series signals (using the AP and ML coordinates of the COP) were filtered using a second-order zero phase Butterworth low-pass digital filter with a cut-off frequency of 10 Hz and an in-house program was used to compute 36 summary measures [2] (Table 3.1 provides brief definitions).

Table 3.1: List of abbreviations (alphabetical order) used to describe the COP summary measures

COP summary measure	Description (units)
Area_CC	95% confidence circle area (mm^2)
Area_CE	95% confidence ellipse area (mm^2)
Area_SW	Sway area (mm^2/s)
CFREQ*	Centroidal frequency (Hz)
FD	Fractal dimension (unitless)
FD_CC	Fractal dimension based on the 95% confidence circle (unitless)
FD_CE	Fractal dimension based on the 95% confidence ellipse (unitless)
FREQD*	Frequency dispersion (unitless)
M_95*	95% power frequency (Hz)
MDIST*	Mean distance (mm)
MPF*	50% power frequency or Median power frequency (Hz)
MFREQ*	Mean frequency (Hz)
MVELO*	Mean velocity (mm/s)
POWER*	Total power (unitless)
RANGE*	Maximum distance between any two points (mm)
RDIST*	RMS distance (mm)

* These measures are computed based on the resultant distance (RD) time series (i.e., the vector distance from the mean COP to each pair of points in the AP and ML time series). Measures are also computed based on the AP time series, and similarly the ML time series.

3.3.4 Statistical analyses

The generalizability theory (G theory) [22] provided a framework to estimate the reliability of the COP summary measures. This modern test theory consists of two parts: the generalizability (G-) study and decision (D-) study. The G-study estimates the various sources of measurement error contributing to the variability in the subjects' values. In the present study, a fully crossed 12 × 2 × 4 (Subject × Day × Trial), two-facet, random effects repeated measures analysis of variance (ANOVA) design was used to represent the conditions of measurement (universe). Thus, the results of the ANOVA were used to obtain the variances attributed to the subjects (σ_S^2), the systematic errors related to the day (σ_D^2) and trial (σ_T^2), and the interactions associated between different sources of variance ($\sigma_{SD}^2, \sigma_{ST}^2, \sigma_{DT}^2, \sigma_{SDT}^2$).

To facilitate interpretation of the results, the proportions of variance (relative to the total variance) attributed to each of these sources of variance were calculated. When performing these calculations, any negative variance components obtained were set to zero [22], which was then used for subsequent calculations involving these variance components.

The subsequent D-study provides the data used to make decisions about the measurement protocol. It estimates the reliability of the observed values corresponding to any study design other than the one used to perform the G-study. The facets considered in these simulations are limited to the one planned for the G-study, but for each facet (Day and Trial), the number of levels to be simulated (e.g., Day = 1, 2, 3 ...) is unlimited.

The sources of variance (from the preceding G-study) were used to calculate the index of dependability (ID or ϕ) and the standard error of measurement (SEM):

$$\phi = \frac{\sigma_S^2}{\sigma_S^2 + \frac{\sigma_D^2}{n_D} + \frac{\sigma_T^2}{n_T} + \frac{\sigma_{SD}^2}{n_D} + \frac{\sigma_{ST}^2}{n_T} + \frac{\sigma_{DT}^2}{n_D n_T} + \frac{\sigma_{SDT}^2}{n_D n_T}} \qquad (1)$$

$$SEM = \sqrt{\frac{\sigma_D^2}{n_D} + \frac{\sigma_T^2}{n_T} + \frac{\sigma_{SD}^2}{n_D} + \frac{\sigma_{ST}^2}{n_T} + \frac{\sigma_{DT}^2}{n_D n_T} + \frac{\sigma_{SDT}^2}{n_D n_T}} \quad (2)$$

where n_T and n_D are, respectively, the number of trials and days averaged when different D-studies are planned. It can be seen from equations (1) and (2) that increasing n_T and n_D will effectively improve the reliability (ID increases and SEM decreases). To determine the effect of different measurement strategies to increase the reliability results, D-studies were simulated where trial facets varied up to 10 trials (i.e., $n_T = 1$ to 10 trials) and day facets varied across 2 days (i.e., n_D=1 or 2 days). The index of dependability, corresponding to the proportion of variance explained by the Subject factor, is analogous to the frequently used intra-class correlation coefficient (ICC) [23]. Like the ICC, the ID ranges between the values 0 (no reliability) and 1 (perfect reliability), thus, interpreted the same way as the ICC: < 0.40 – poor, 0.40-0.75 – moderate and > 0.75 – excellent [24]. The SEM provides an indication of the absolute reliability of the measure (same units of measurement). To better judge the relative importance of SEM values, they were expressed as a percentage (%SEM) of the grand mean calculated across days, which is analogous to the coefficient of variation (CV). This choice has the disadvantage of giving large %SEM values when the mean is around zero. However, in such situations, IDs give a better indication of reliability.

3.4 Results

The mean (SD) values across all four trials and two days are presented in Table 3.2. Where applicable, only measures computed on the AP and ML (rather than the resultant) time series are presented as this is the convention most commonly reported in the literature.

3.4.1 Generalizability Study

The magnitude of the variance components (expressed as a percentage of the total variance) is presented in Table 3.2. On average, the σ_S^2 was 46.5% (range: 3% [POWER_AP] to 74.9% [CFREQ_ML]) and 40.1% (range: 2.2% [POWER_ML] to 71.6% [CFREQ_ML]) in the EO and EC conditions, respectively, across all COP summary measures computed. The proportion of variance attributed to the subject (σ_S^2) corresponds to the ID (or ICC) when n_T and n_D are equal to 1. Therefore, using established criteria [24], the reliability of the COP measures (using 60 s trials) was qualified as poor to moderate.

The contribution of the day facet (σ_D^2) variances was less than 2.0 % (EO) and 3.3% (EC). Likewise, the variance estimates for the trial facet (σ_T^2) was minimal, contributing less than 1% (EO) and 0.9% (EC) to the overall measurement error (not presented in table 2). These small values indicate that negligible between- and within-day systematic errors were present in the current study design.

The error variance related to the day × trial interaction (σ_{DT}^2) was, on average, small compared to the total variance (not presented in Table 3.2) across all summary measures and conditions. The average σ_{DT}^2 in EC was slightly higher than EO (4.5% versus 1.8%) across all summary measures. This was due to the fact that with EC, FREQD_ML reached 20.3% and seven other summary measures reached between 11% and 16.3%; the remaining did not exceed 9.8%. With EO, all summary measures showed σ^2_{DT} smaller than 11%.

The contribution of σ^2_{SD} was larger, on average, in EO (11.4%) as compared to EC (3.5%) across all summary measures (Table 3.2). In EO, the σ^2_{SD} for 56% (20/36) of the summary measures was greater than 10% (range: 10.5% - 23.4%). In EC, however, σ^2_{SD} of all summary measures was below 6%, with the exception of only 3 summary measures exceeding 10%. The σ^2_{ST} was also small, contributing, on average, to 3.1% and 2.5% of the overall variance for the EO and EC conditions, respectively. There were, however, a few exceptions (Table 3.2).

Apart from σ^2_S, the largest proportion of measurement variability was attributed to σ^2_{SDT}, the variance attributed to the highest order interaction, combined with the residual error (Subject × Day × Trial). This contributes an average of 36% across all the summary measures in EO (range 16.2% [MVELO_ML] to 60.2% [POWER_AP]) and an average of 48% in EC (range 23.8% [CFREQ_ML] to 78.7% [RANGE], not shown in Table 3.2).

Table 3.2: Mean, standard deviation (SD), and relative magnitude (%) of the variance components from the G-study for the COP time-domain and frequency-domain summary measures* for EO and EC

	Eyes Open (EO)						Eyes Closed (EC)					
	Mean	SD	σ^2_S	σ^2_{ST}	σ^2_{SD}	σ^2_{STD}	Mean	SD	σ^2_S	σ^2_{ST}	σ^2_{SD}	σ^2_{STD}
Time Domain Measures												
MDIST_AP	2.8	1.0	34	10	18	39	3.0	1.2	46	0	1	52
MDIST_ML	3.7	1.6	43	0	6	48	4.2	1.7	39	7	5	37
RDIST_AP	3.4	1.2	40	12	15	33	3.8	1.5	46	0	2	50
RDIST_ML	4.6	1.9	45	0	8	44	5.3	2.1	43	5	6	40
RANGE_AP	18.5	6.4	55	7	7	30	22.7	18.6	19	1	0	78
RANGE_ML	23.9	7.8	48	0	8	40	29.5	11.9	36	0	0	61
MVELO_AP	7.1	1.0	44	0	8	40	8.1	1.3	32	10	7	38
MVELO_ML	8.7	1.4	46	13	21	16	10.6	1.8	41	0	0	46
AREA_CC	328	247	39	0	22	38	432	345	37	0	1	58
AREA_CE	467	337	40	0	22	36	597	455	43	0	0	52
AREA_SW	20.9	8.2	55	0	5	38	27.4	12.4	38	0	0	50
MFREQ_AP	0.5	0.2	36	14	14	36	0.5	0.2	49	0	6	42
MFREQ_ML	0.5	0.2	43	0	7	45	0.5	0.2	55	9	2	31
FD	1.6	0.1	37	0	14	46	1.6	0.1	38	0	0	59
FD_CC	1.8	0.1	50	3	20	24	1.8	0.1	59	5	5	29
FD_CE	1.7	0.1	52	4	20	23	1.7	0.1	63	5	6	24
Frequency Domain Measures												
POWER_AP	36.4	16.2	3	25	12	60	43.4	15.8	46	0	16	70
POWER_ML	25.9	14.4	16	0	14	58	37.5	14.4	40	1	23	69
MPF_AP	0.32	0.07	46	0	7	47	0.33	0.09	46	6	0	47
MPF_ML	0.36	0.07	53	0	0	47	0.37	0.10	43	1	0	46
M_95_AP	1.15	0.20	57	2	9	29	1.17	0.29	19	0	0	67
M_95_ML	1.24	0.27	74	0	4	21	1.16	0.24	36	0	0	32
CFREQ_AP	0.63	0.10	69	0	5	26	0.63	0.13	32	0	0	54
CFREQ_ML	0.69	0.13	75	0	1	22	0.64	0.13	41	0	0	24
FREQD_AP	0.68	0.04	45	0	14	40	0.67	0.04	43	8	0	54
FREQD_ML	0.67	0.04	49	0	20	48	0.64	0.05	38	0	3	46

* The measures based on the resultant distance time series (i.e., the vector distance from the mean COP to each pair of points in the in the AP and ML time series) were computed, however not presented in this table. Only measures computed based on the AP and ML time series are presented.

3.4.2 Decision Study

The general trend resulting from the simulation studies for all summary measures was that as the number of trials averaged increased and as the number of days over which these trials were averaged, so did reliability (i.e., higher indices of dependability and lower standard error of measurement), as illustrated in Figure 3.1 for a selection of summary measures.

In general, 47 % (17/36) and 78% (28/36) of the summary measures reached excellent reliability (ICC >0.75) by averaging at least 7 (or less) trials over one day for EO and EC, respectively (Table 3.3; not all summary measures presented). For measures requiring more than 7 trials, the SEM was less than 20%, with only a few exceptions (see Table 3.3), and fewer trials are generally needed if averaged over two days.

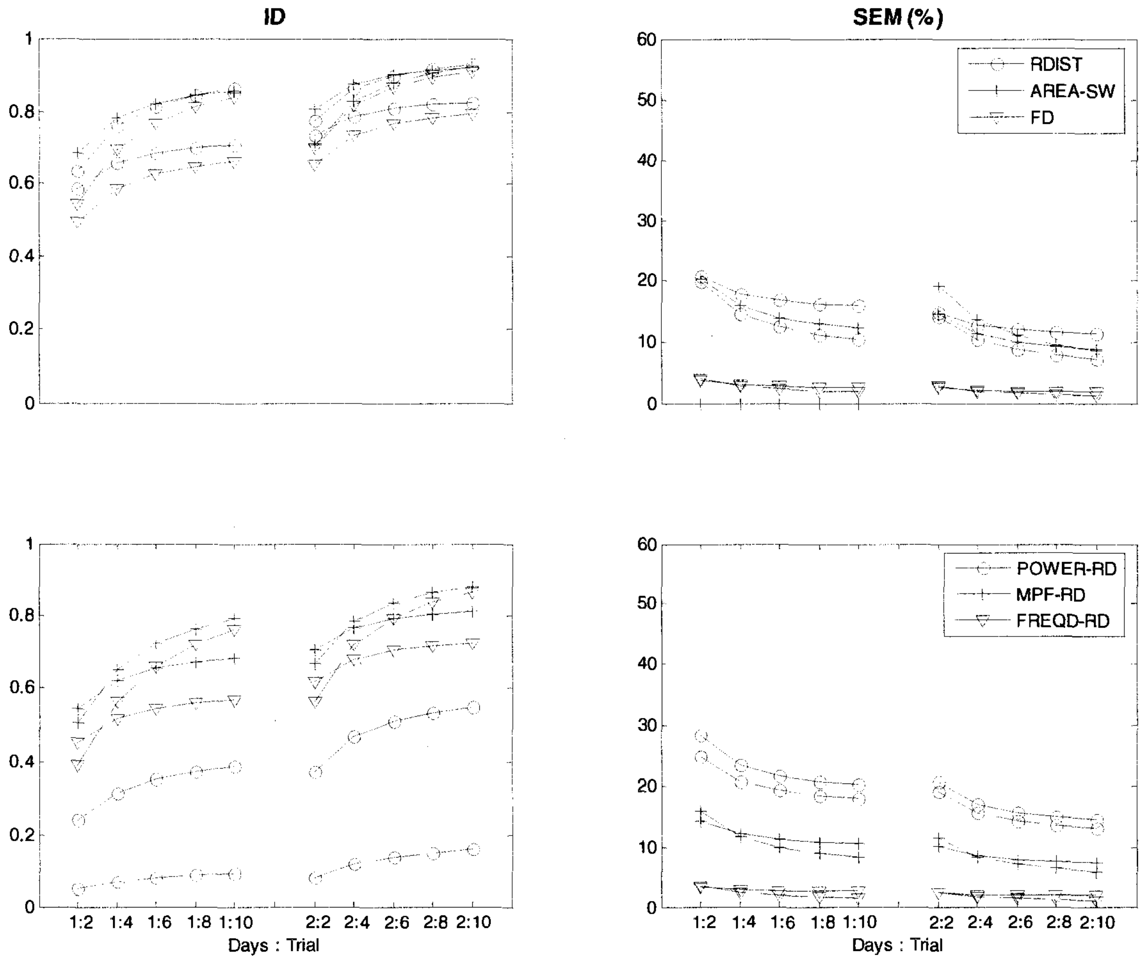

Figure 3.1: Reliability statistics (index of dependability [ID] and standard error of measurement [SEM]) as a function of the number of trials (2, 4, 6, 8, 10) averaged over the number of days (1,2) obtained from the decision (D-) study measurement strategies for selected COP time-domain (top plots) and frequency-domain (bottom plots) summary measures. Solid lines (EO). Dashed lines (EC). Left plots (ID). Right plots (%SEM).

Table 3.3: Number of trials needed to reach excellent reliability (with corresponding ϕ and %SEM values)

	Eyes Open (EO)						Eyes Closed (EC)					
	1 Day			2 Days			1 Day			2 Days		
	# trials	ϕ	SEM %	# trials	ϕ	SEM %	# trials	ϕ	SEM %	# trials	ϕ	SEM %
Time Domain Measures												
MDIST_AP	> 10	-	17.8	10	0.75	12.9	4	0.77	15.8	2	0.77	15.7
MDIST_ML	6	0.75	17.5	3	0.79	15.7	7	0.75	15.4	3	0.75	15.3
RDIST_AP	> 10	-	16.1	5	0.75	13.2	4	0.75	16.8	2	0.77	16.1
RDIST_ML	6	0.75	17.0	3	0.79	14.9	6	0.77	14.9	3	0.79	14.1
RANGE_AP	4	0.77	14.6	2	0.79	13.7	> 10	-	27.3	8	0.75	21.2
RANGE_ML	5	0.75	13.8	2	0.75	13.5	7	0.76	15.5	3	0.76	15.5
MVELO_AP	7	0.75	5.6	3	0.79	5.0	> 10	-	6.0	5	0.76	5.4
MVELO_ML	> 10	-	8.1	5	0.75	6.4	5	0.77	6.7	3	0.81	6.1
AREA_CC	> 10	-	39.5	9	0.75	28.3	6	0.75	29.5	3	0.77	28.0
AREA_CE	> 10	-	38.9	8	0.75	28.1	4	0.76	30.0	2	0.76	30.0
AREA_SW	3	0.75	17.5	2	0.81	14.6	5	0.75	17.1	3	0.79	15.6
MFREQ_AP	> 10	-	14.5	6	0.75	11.7	4	0.75	13.3	2	0.77	12.3
MFREQ_ML	7	0.75	12.8	3	0.77	12.1	3	0.78	12.9	2	0.79	12.3
FD	> 10	-	2.6	5	0.76	2.1	6	0.77	2.3	3	0.77	2.3
FD_CC	> 10	-	2.9	3	0.76	2.4	3	0.77	2.5	3	0.86	1.9
FD_CE	> 10	-	2.7	3	0.77	2.3	2	0.75	2.7	3	0.87	1.7
Frequency Domain Measures												
POWER_AP	> 10	-	21.0	> 10	-	15.7	> 10	-	18.0	> 10	-	12.8
POWER_ML	> 10	-	27.4	> 10	-	19.4	> 10	-	21.7	> 10	-	15.5
MPF_AP	6	0.76	9.3	2	0.75	9.6	4	0.76	10.2	3	0.81	9.0
MPF_ML	3	0.77	10.9	2	0.82	9.4	4	0.77	11.5	2	0.78	11.2
M_95_AP	4	0.76	7.4	2	0.80	6.7	7	0.77	8.6	4	0.79	8.1
M_95_ML	1	0.75	11.4	1	0.85	8.2	4	0.77	10.3	2	0.78	10.2
CFREQ_AP	2	0.80	7.3	1	0.82	6.8	4	0.76	8.5	2	0.76	8.5
CFREQ_ML	1	0.75	10.4	1	0.86	7.4	2	0.82	8.6	1	0.83	8.3
FREQD_AP	> 10	-	2.9	3	0.76	2.5	6	0.75	8.5	4	0.78	8.5
FREQD_ML	4	0.78	2.5	2	0.79	2.4	9	0.75	2.7	4	0.76	2.7

3.5 Discussion

The purpose of this present investigation was to assess the reliability of a large set of COP summary measures [2] using the generalizability theory [22]. Unlike the classical test theory, the generalizability analysis allows researchers and clinicians to estimate both the magnitude and relative contribution of different sources of measurement error. Being aware of these errors would allow for the investigators (i.e., researcher, clinician) to correct for them. Furthermore, investigators would be able to make decisions as to different measurement strategies that provide them with optimal reliability depending on their study population and budget.

Before continuing with the discussion, the authors acknowledge the small sample size (n=12) as a limitation to the study, which could affect the stability of the estimates of the variance components. A larger sample size, however, would increase research costs not only financially but also in time commitment. In previous studies investigating the reliability of COP measures the number of participants varied from as few as seven to 49 participants [10, 13, 16, 25]. Nonetheless, the number of subjects who participated in the current study is in the same range (n=15) as another study using the generalizability theory as a framework [21], but the current study is the second to address between-day variability. The only other study to investigate between-day variability assessed the optimal test duration [17].

3.5.1 Generalizability Study

The proportion of variance attributed to the subject, corresponding to the ICC when both n_T and n_D are equal to 1, showed the reliability of the COP summary measures ranging from poor to moderate. The use of vision did not systematically improve reliability. However, the D-study results revealed quite different results (to be discussed later). For 56% of the variables (20/36), reliability was higher for EO than EC.

The strategy used to compensate for the lack of visual information may account for such differences.

In a reliability study, the possibility of the presence of any systematic errors must be verified. In this investigation the variance explained by σ_D^2 and σ_T^2 was low (< 4%), indicating negligible systematic effects. This demonstrates that the protocol was successful at avoiding subjects from becoming fatigued (that would have affected σ_T^2) even though no rest periods were allocated between trials. Previous studies [16,25] have given as much as five minutes of rest between trials to avoid fatigue. It also suggests that this task is not prone to motor learning within and between days.

The reliability of COP measures has previously been addressed. Due to different measurement protocols (i.e., different sampling durations, feet stances, EO versus EC); it is difficult to compare our results with those published elsewhere. Even though force plate measurements are an accepted method for evaluating postural balance, there is a lack of a standardized measurement protocol. Furthermore, the reliability of many of the summary measures that we have presented has not yet been reported in previous literature. The index used to measure reliability also varies from one study to another; some report the ICC [10,13,25,26], others report the coefficient of variation (CV) [18,27] or both [14]. The %SEM is analogous to the CV. Neither the ICC nor the SEM is a surrogate measure for the other. In fact, ICCs indicate the potential to discriminate between subjects giving an idea of the diagnostic value of a measure (e.g., between-subjects designs) while SEMs shows the capacity to detect changes over time (e.g., the effect of a rehabilitation program, within-subjects designs). Several factors may influence the magnitude of the variance between subjects as well as the error variance (i.e., study design, study population, therapeutic intervention, etc) and hence, may explain why differences in reliability are reported in the literature.

Depending on the summary measures, our reliability results varied from those that have been previously published [16,21] (Table 3.4). For example, Lafond et al. (2004) reported excellent ICC values for COP velocity whereas we had moderate values. However, for MPF and MedMP we reported moderate reliability score whereas they had poor reliability scores. Differences could have been due to the different population studied (elderly versus young individuals). Doyle et al. (2007) report moderate reliability values ranging from 0.31 (AREA_CE) to 0.63 (COP velocity) for EO, and 0.28 (standard deviation – equivalent to our RMS – AP) to 0.62 (COP velocity) for EC (Table 3.4). We obtained slightly higher reliability values (except for COP velocity). Given that both study populations and protocols were very similar, the slight differences in reliability values are difficult to explain.

The components of the measurement variance revealed that the majority of the measurement error was random. A large contribution to the variability of the measurement was the highest order interaction (σ^2_{SDT}), which contains the unexplained random variance and/or error variance attributed to facets not identified in the study.

Table 3.4: Comparison of the reliability for some traditional COP summary measures based on one 60-s trial (on one day)

COP Summary Measure	Direction	Lafond et al. (2004)		Doyle et al. (2007)		Current study	
		EO	EC	EO	EC	EO	EC
RMS	AP	0.52	-	0.38	0.28	0.40	0.46
	ML	0.61	-	0.41	0.32	0.45	0.43
Velocity	RD	-	-	0.63	0.62	0.53	0.44
	AP	0.77	-	-	-	0.44	0.32
	ML	0.90	-	-	-	0.46	0.41
RANGE	AP	0.38	-	-	-	0.55	0.19
	ML	0.57	-	-	-	0.48	0.36
MFREQ	AL	0.09	-	-	-	0.36	0.49
	ML	0.20	-	-	-	0.43	0.55
MPF	AL	0.02	-	-	-	0.46	0.46
	ML	0.24	-	-	-	0.53	0.43
AREA_SW	-	0.47	-	-	-	0.55	0.38
AREA_CE	-	-	-	0.31	0.30	0.40	0.43

3.5.2 Decision Study

In general, 47% (17/36) and 78% (28/36) of the summary reached excellent reliability (ICC > 0.75) by averaging 7 trials over one day for EO and EC, respectively. For both conditions, although reliability increased as more trials were averaged within the same test day, reliability was increased more substantially by averaging trials across days. These results are consistent with a much higher percentage of Subject × Day variance (σ^2_{SD}) than Subject × Trial variance (σ^2_{ST}), which reflects that the summary measures, for each subject, was more affected by between-day (Day factor) than within-day (Trial factor) sources of error, relative to other subjects. However, this would not necessarily be the most practical situation especially for evaluations in a clinical situation. From a practical standpoint, averaging trials from one day would be more ideal than two days. Caution should be taken if one increases the number of trials on a single testing session. Although the results from this study demonstrated minimal systematic error due to the Trial factor, we cannot speculate on the effect on subject boredom or fatigue if more than four trials are performed.

As with the D-study of Doyle et al. (2007), we too found that fewer trials are needed to reach acceptable reliability with EC than with EO. Thus, increasing the number of trials appears to be a good strategy to improve reliability especially with EC. A note should be addressed regarding the use of the ICC as an index of reliability. Little variability among the subjects will lead to lower ICC values. The homogeneity of our subject sample (young and healthy subjects) could reduce the variability among the subjects and consequently lead to lower reliabilities. Despite the ICCs, it should be kept in mind that the SEM is also of importance when evaluating an individual subject/patient. If repeated measures are made and the measurement of changes (e.g. resulting from a treatment) is the priority, then SEM is the index of interest. Thus, even though some of the summary measures required more than 10 trials to obtain an ICC > 0.75, the corresponding SEM values for these measures (with some exceptions) were relatively low (<20%) (Table 3.3).

D-studies play an important role in the design of basic and clinical science experiments. There is an obvious tradeoff between achieving a desired level of reproducibility and having enough resources to satisfy the time and cost for the number of sessions and/or trials required for the reproducibility. Furthermore, there are limits to what is expected of human participants in terms of time commitment. If measurements are taken in a rehabilitation setting, patients may not be able to tolerate repeated measurements over single or multiple sessions. Consequently, all these factors must be taken into account by the investigator performing these measurements.

3.6 Conclusion

The present study estimated the reliability of a large set of COP summary measures among a population of healthy male subjects. Performing one 60-s trial on one day of standing balance yielded poor to moderate reliability depending on the measure. The reliability could be improved more efficiently by averaging measurements between-days than by increasing the number of trials during one day. However, for the majority of the summary measures, acceptable reliability can be achieved when at least 7 trials are averaged during the same testing day.

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CHAPTER 4

SUDDEN LOADING PERTURBATION TO DETERMINE THE REFLEX RESPONSE OF DIFFERENT BACK MUSCLES: A RELIABILITY STUDY

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"Muscle and Nerve"

4.1 Abstract

Adequate reflex responses of the lumbar muscles are important in maintaining appropriate spinal stability. The study aimed to estimate the reliability of reflex response variables, elicited through a sudden loading perturbation, so that the main sources of variability could be identified and to estimate the most appropriate measurement strategies to obtain more reliable measures. Back muscles electromyography (EMG) and trunk kinematics were recorded in 15 healthy males during anteriorly-directed sudden loading perturbations applied to the trunk in a no preload and a preload condition, performed on two separate occasions within the same day and then repeated on a second day. Measures of EMG reflex latency and amplitude, as well as of trunk kinematics were obtained. The generalizability theory was used as a framework to estimate the magnitude of the different variance components (Subject, Day, Test, Trial and all interactions) and the reliability of the measures corresponding to various simulations of different measurement strategies. Reliability of the different variables ranged from poor to moderate, with intra-class correlation coefficients ranging between 0 and 0.62. Averaging the scores across homologous muscles and several trials were shown as practical strategies to achieve more acceptable reliability. This study showed that the reflex response of back muscles is inherently variable and that a large measurement effort is necessary to obtain reliable and consequently, valid and responsive estimations of this neuromuscular function.

KEYWORDS: Sudden loading; Back muscles; Reflex response; Reliability; Measurement strategies; EMG onset detection methods

4.2 Introduction

Studies on sudden loading of the trunk are important from both an occupational health standpoint and in the investigation of spinal stability. From the occupational health view, large spinal muscle forces are often needed to maintain balance when the trunk has been perturbed in a certain direction[9, 17], which produces large internal loading[17]. Within the context of lumbar stability, these loads also challenge the stability of the spinal system. Muscle forces introduce a certain amount of stiffness and postural reflexes alter muscular activity so that balance is maintained[21, 22]. Adequate reflex responses of the lumbar muscles are thus important in maintaining spinal stability. Consequently, well-standardized measurement protocols must be developed to study this phenomenon properly.

Along with intrinsic muscle stiffness, reflex responses are a necessary component in the stabilizing control of spinal stability [20]. Non-invasive and indirect methods of measuring the reflex responses have been used with either a sudden loading paradigm[14, 32, 37] or a sudden unloading (quick release) paradigm[5, 23]. Measurement of the reflex response through muscle elongation via joint angular displacement (through the aforementioned sudden (un)loading paradigms) is expected to be more representative of everyday activities. However, the origin of the reflex response, whether it be the stretch reflex, or from the ligament, facet capsule or the discs, using this method is unknown. Using sudden (un)loading paradigm, the control of the perturbation (i.e., amplitude, velocity, acceleration) is difficult. This has been performed in measuring lower-leg muscles' reflexes by controlled ankle joint perturbation[13], however this has not been applied to the lumbar joints. Additionally, the control of the muscle state before the perturbation is difficult and should also be standardized. The pre-tensioning or not of the investigated muscles with the use of a preload or not[3], as well as the load anticipation of the participant[35] (i.e., an increase of muscle activation with load anticipation) will influence the amount of joint stiffness.

To determine the reflex response, more specifically the EMG reflex latency, the precise determination of the EMG onset is required. A simple and common method for determining event detection is with off-line visual inspection. The nature of EMG signals is very complex, thus visual onset determination tends to be inconsistent because of observer detection error. In addition to being very subjective, the criteria used with a manual (visual) technique for onset detection is not often described by the researchers. Therefore, the reliability of visual inspection could be questioned. To address this problem, several automatic and computerized methods for event detection have been developed even though there is little consensus as to which is the most appropriate method. A more in-depth comparison of some computerized methods should be performed with regard to reliability.

Documentation of the reliability of variables used in sudden loading studies is sparse, at best, in the literature. The reproducibility of a sudden loading test repeated over 10 trials was investigated[26]. Through an analysis of variance, the investigators found that the reaction time (latency) of the first trial was significantly longer than from trials 3-10. In this study, reliability coefficients were not calculated and visual determination of the EMG onset was used. Hermann et al. (2006) measured spinal muscle reflexes using anterior-perturbations that were applied, while subjects were standing quietly, using a pendulum suspended by the ceiling. Three trials before and after a fatiguing task were performed. The reported intra-class correlation coefficient (ICC) for reflex delay and amplitude were 0.41 and 0.61, respectively. Unfortunately, we do not know how many trials would be necessary to increase these ICC values to a more acceptable level of reliability.

The purpose of this study was to compare the reliability of 1) different computerized EMG reflex latency and amplitude estimates and 2) of using a pre-load or not with the sudden loading paradigm. Measurement strategies for improving the reliability of these measures were also investigated.

4.3 Methods

4.3.1 Subjects

Fifteen healthy males were recruited to participate. Participant mean (SD) age, height and weight were 26.1 (4.7) years, 1.8 (0.1) m, and 74.9 (12.8) kg, respectively. Subjects were excluded if they had a systemic, degenerative or neurological disease, a musculoskeletal problem, low-back pain lasting more than one week or requiring medical attention during the 12 months prior to participation in the study or responded positively to the revised Physical Activity Readiness Questionnaire[33]. The subjects were informed of the experimental and potential risks and provided written consent prior to their participation. Each subject read and signed an informed consent form approved by the Laurentian university Research Ethics Board.

4.3.2 Task and procedure

The reflex response of the back muscles was measured using a sudden loading apparatus (Figure 4.1) designed to give an anteriorly directed perturbation of the trunk[14]. Before each trial, the subject was asked to position his trunk in the same reference position (zero position when the trunk was upright and still), using a visual feedback from a potentiometer.

Two pre-load conditions were used: no pre-load (NP) and a pre-load equivalent to 15% of the trunk mass (or less than 7.5% of the total body mass[7]) (PRE15). This pre-load helps to pre-activate the back muscles in a standardized manner from trial to trial. To minimize the pre-activation of the abdominal muscles, visual feedback of the two abdominal muscles was displayed for the subject.

The load (pre-load and added load) was hidden by a metal screen to remove any visual clues to eliminate any possible effects of anticipation. The added load was released randomly between 5 and 15 seconds after the trial had started. This added load, equivalent to 35% of the trunk mass, was large enough to minimally solicit the back muscles though not large enough to inhibit the reflex response. Lastly, subjects were instructed to stop their trunk displacement as soon as the added load had been released, but to avoid overreacting by extending the trunk backward.

One data collection session required that the participant perform a maximum of 25 sudden loading perturbations over one testing day. After the subject was stabilized in the sudden loading apparatus, two to five anteriorly-directed perturbations were performed to become accustomed to the task. Five sudden loading trials were then performed for both pre-load conditions (NP and PRE15) with approximately a 30 s rest period between each trial. The NP and PRE15 conditions were presented in a counterbalanced design across subjects. After these first 10 trials were performed, the participant came out of the apparatus and was given a rest period where he walked around the laboratory for 15 minutes. Then, to assess whether repositioning the subject in the apparatus could add variance in the results, they were stabilized back into the apparatus and performed the same procedure (Test 2). To assess the possible effect of between-day sources of variance (i.e., learning, EMG electrode repositioning), each participant returned to the laboratory, between 2 and 7 days after the first visit, and performed the same procedure with the sequence of conditions being presented as in the first visit.

4.3.3 Electromyography

During the sudden loading test, muscle activation levels from eight sites were measured with surface electromyography (EMG), using differential pre-amplified (gain: 1000, band-pass filter: 20-450 Hz) active surface electrodes (Model DE-2.3, Delsys Inc., Wellesley, MA, USA) composed of two silver bars (spaced 10 mm, 1 mm wide). The raw EMG signals were analog to digital converted at a sampling rate of 1024 Hz (12-bit, PCI-6071E, National Instrument, Austin, TX, USA) and stored on a hard disk for later analysis. After the skin at the electrode sites were shaved, gently abraded and cleaned with alcohol, the electrodes were positioned bilaterally on the longissimus at the level of L1 (LONG-L1-L and LONG-L1-R), iliocostalis lumborum at L3 (ILIO-L3-L and ILIO-L3-R) and multifidus at L5 (MULT-L5-L and MULT-L5-R) following the recommendations of De Foa et al (1989). The difficulty in capturing the multifidus muscle with surface electrodes[31] is acknowledged and therefore the validity of the electromyographic signal was assigned to the landmarked location rather than to the multifidus muscle itself. Additional electrodes were positioned on the right rectus abdominus and right external obliques as per McGill (1991). To ensure the same placement of the electrodes for the back muscles from day to day, a template using visible anatomical landmarks was used. A reference snap-on type surface electrode (Medi-Trace model, Graphic Controls Canada Limited, Gananoque, Ontario, Canada) was positioned on the spinous process at the C7 level.

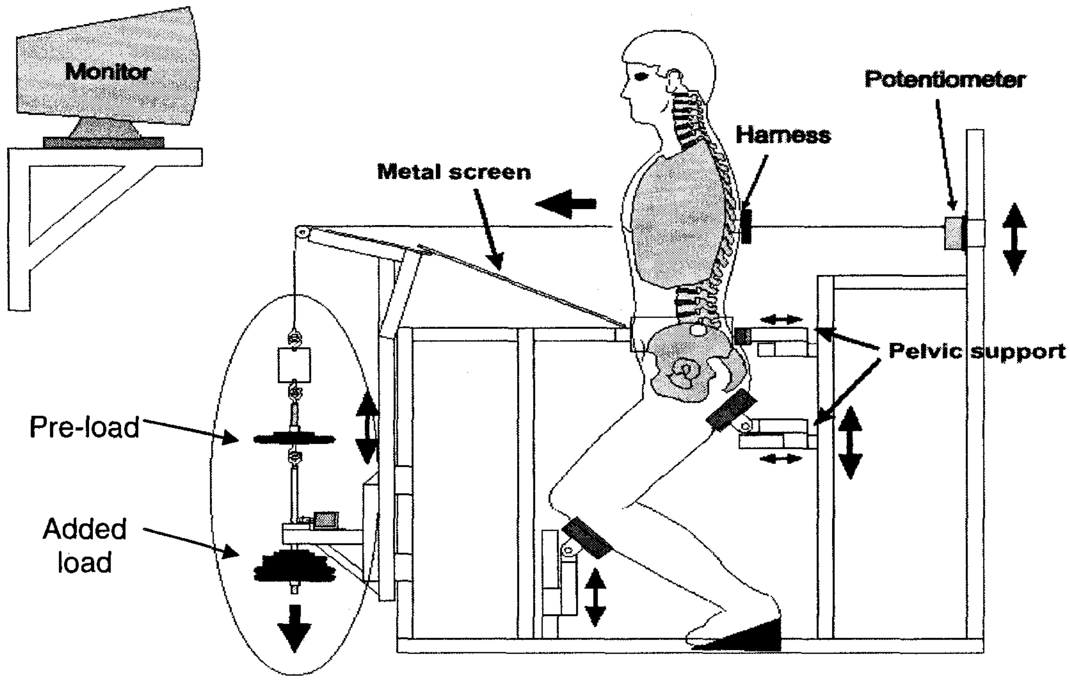

Figure 4.1: Sudden loading apparatus. This apparatus allows for the stabilization of the subject's lower extremities and pelvis. The sudden load was applied via a cable connected to a load cell and attached (at the level of T8) to a harness adjusted on the subject's chest and shoulders. This load, initially held by an electro-magnetic release mechanism, was released from a minimal height of approximately 1 cm to minimize ballistic loading effects, thus assuring the safety of the test.

4.3.4 Kinematics

Trunk displacement was measured via a cable that was attached to the back of a harness worn by the participant and connected to a potentiometer (Model P-30AiT A159, Patriot Sensor & Controls Corporation, Rayelco Linear Motion Transducer, Simi Valley, CA, USA). The cable was adjusted so that it was parallel to the ground at the level of T8.

4.3.5 Data processing

All data processing and data reduction were performed using MATLAB (Version 7.0, The Mathworks, Natick, MA, USA). The following sections describe how the different outcome variables were computed.

4.3.5.1 Assessment of electromyographic reflex response

Post-processing of all EMG signals involved several steps. Signals were first filtered using a wavelet method to remove ECG artefact[2]. Briefly, the algorithm chose the most appropriate wavelet transformation among 15 possible wavelets (Daubechies 4-10, Meyer, Coiflet 2-6, Symlet 4, 6-8) that would give the EMG signal a standard deviation closest to the standard deviation of the signal without ECG. A notch filter was then applied to eliminate possible 60-Hz electrical noise and its harmonics (up to 420 Hz).

Computation of the reflex latency

The reflex latency was defined from the beginning of trunk movement to the onset of the EMG response (Figure 4.2). The 5 s preceding the force perturbation was used as the common EMG reference signal (baseline) to all methods. A window of 250 ms after the force perturbation was used to search for an EMG response. The EMG onset was determined using three different computer-based automated methods.

The first method was the Shewhart (SHEWHART) method[12]. The signal was dual-pass Butterworth filtered (effective 6^{th}-order 50 Hz low-pass cut-off). The processed EMG signal was assessed using a 25-ms sliding window. If the signal exceeded a threshold of two standard deviations (SDs) above the baseline mean, then a muscle response to the force perturbation was considered to have occurred.

The second method implemented was the approximated generalized likelihood-ratio (AGLR) model-based algorithm that uses the log-likelihood ratio in order to estimate the probability of a portion of an EMG signal to pertain to a certain reaction variance in comparison to the variance at rest[27, 28]. Essentially, segments of signals were compared to see the likelihood of them being statistically different using a pre-defined likelihood threshold set at 75% of the maximum of the function.

The final method used a wavelets method (WAVELET). The underlying model of the EMG signal was represented by an uncorrupted signal added to a gaussian white noise of level σ:

$$signal(n) = f(n) + \sigma \cdot e(n).$$

Noise was first removed from the input signal using Daubechies wavelet (dB1) with a soft threshold. The threshold was approximated by the square root of two times the logarithm of n, then rescaled using the median of the absolute value of the detailed decomposition coefficient at the first level:

$$Threshold = \sqrt{2 * \log(n)} * median(abs(c)) * 0.6745$$

where n is the number of samples and c the detailed coefficient at level 1. The approximation coefficients were kept unchanged. Only the detailed coefficients of levels

1, 2, and 3 are filtered and used to determine the reflex time. With the new coefficients, the maximum standard deviation is calculated on each detailed level during the rest period. A sliding window of 10 ms was used to calculate the standard deviation each window at a time. A two-standard deviation criterion was used. The reaction times at each frequency band were found and the lowest time was taken as the true reaction time. For all the three methods (SHEWHART, AGLR, WAVELET), reflex latencies <30 ms and >150 ms following trunk movement were assumed to be non-reflexive responses and eliminated from subsequent analysis.

Computation of the reflex amplitude

The reflex amplitude was quantified in two different ways, but only when a reflex latency was detected in the 30-150 ms time interval following trunk movement. The EMG signal was first rectified and dual-pass second-order Butterworth filtered (2^{nd}-order, 25 Hz cut-off frequency). The first method (RatioPeak) was the ratio of the first EMG peak value (after the EMG onset as detected using the SHEWHART method) divided by the EMG signal (250 ms) prior to trunk movement. The second method (RatioArea) was the ratio of the area under the curve from the EMG onset (as detected using the SHEWHART method) to the first EMG peak (i.e., the same peak as the previous method), divided by the area under the curve corresponding to the EMG signal (250 ms) prior to trunk movement. The parameter RatioArea thus is dependent on the reflex amplitude and rise time.

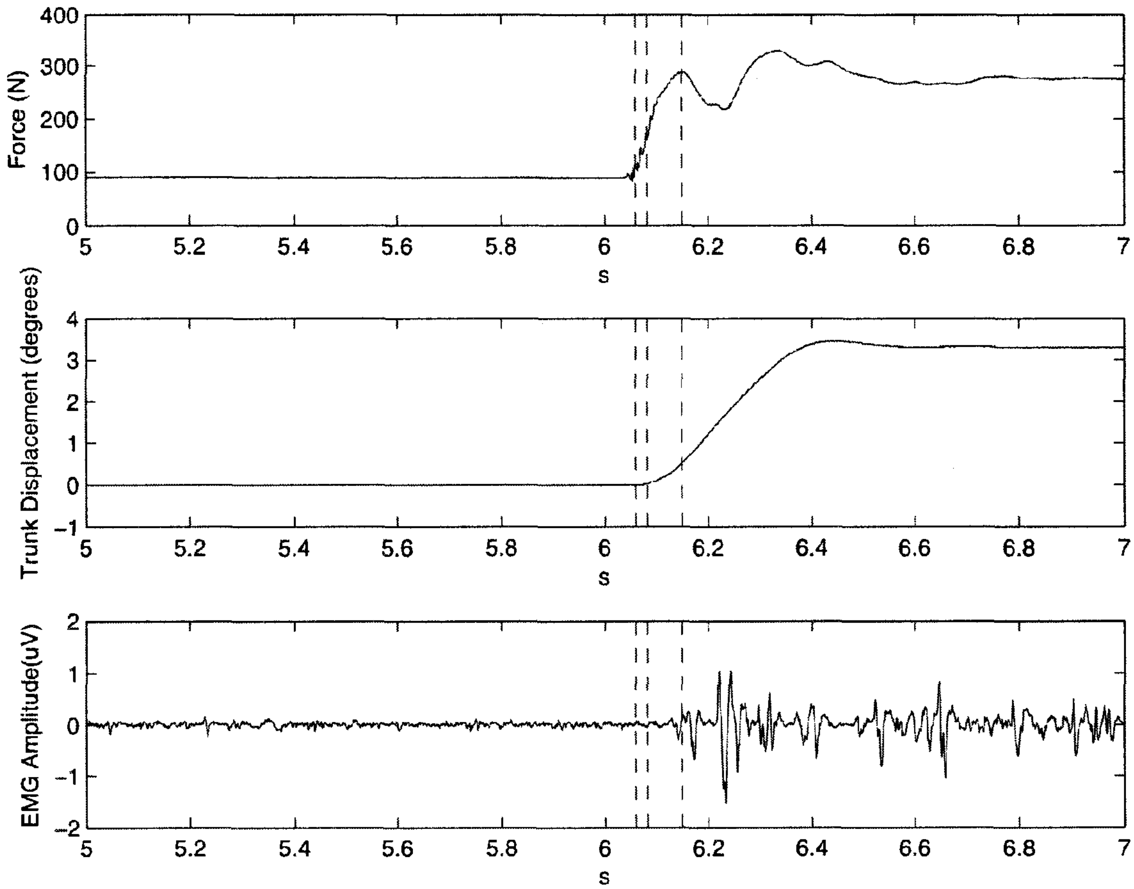

Figure 4.2: Example of a sudden loading trial. The top plot is the force signal measured by the load cell attached to the load. The middle plot represents the potentiometer signal measuring trunk displacement. The bottom plot is an EMG signal from one of the back muscles. The EMG reflex latency (ms) was calculated as the time between the detection of the first movement of the trunk and the moment the EMG onset was determined. The EMG reflex amplitude was calculated as the ratio of the maximal EMG reflex signal (EMGReflex) to the baseline EMG signal (EMGPrePerturbation).

4.3.5.2 Kinematics

Signals were dual-pass filtered with a 2^{nd} order Butterworth filter using a low-pass cut-off frequency of 20 Hz. The detection of the initial movement of the trunk after the sudden load release was determined using a modified version of the log-likelihood method[28], where constraints were added on force time derivative values. This method can be used in a variety of signals in which steep changes in variance are characteristic of event detection. Filtered data were initially in radians, but were subsequently converted to degrees. Angular displacement (degrees) was determined and then maximal and average angular velocity (degrees/s) and maximal angular acceleration (degrees/s^2) were derived from the trunk displacement data.

4.3.6 Statistical analysis

Preliminary analyses showed no statistical difference (T-test, $\alpha = 0.05$) between homologous (left and right) muscles. This justified the use of bilateral averaging of reflex variables to obtain one score at each vertebral level (L5, L3, L1) in some of the following analyses. For each EMG variable and each muscle, a three-way (2 DAY × 2 TEST × 5 TRIAL) analysis of variance (ANOVA) with repeated measures on all factors was used.

The generalizability theory (G Theory)[24] provided a framework to estimate the reliability of the computed EMG and kinematics variables. This modern test theory consists of two parts: the generalizability (G-) study and the decision (D-) study. The G-study estimates the various sources of measurement error contributing to the variability in the participants' values. In the present study, a fully crossed 15 Subjects × 2 Days × 2 Test × 5 Trials, three-facet (Day, Test and Trial), random effects repeated measures analysis of variance (ANOVA) design was used to represent the conditions of measurement (universe of the possible sources of variance). The Test facet refers to the different sets of trials performed during the same day, implying the repositioning of the subject in the apparatus. Thus, the results of the ANOVA were used to obtain the variances attributed to the subjects (σ_S^2), the systematic errors related to the day (σ_D^2), test (σ_{Te}^2), and trial (σ_{Tr}^2), and the interactions associated among the different sources of variance (σ_{SD}^2, σ_{STe}^2, σ_{STr}^2, σ_{DTe}^2, σ_{DTr}^2, σ_{TeTr}^2, σ_{SDTe}^2, σ_{SDTr}^2, σ_{STeTr}^2, σ_{DTeTr}^2, σ_{SDTeTr}^2).

To facilitate the interpretation of the results, the proportions of variance (relative to the total variance) attributed to each of these sources of variance were calculated. When performing these calculations, any negative variance components obtained were set to zero[24], which was then used for subsequent calculations involving these variance components.

The subsequent D-study provides the data used to make decisions about the measurement protocol. It estimates the reliability of the observed values corresponding to any study design other than the one used to perform the G-study. The facets considered in these simulations are limited to the one planned for the G-study, but for each facet (Day, Test, and Trial), the number of levels to be simulated (e.g., Day = 1, 2, 3, …) is unlimited.

The sources of variance (from the preceding G-study) were used to calculate the index of dependability (ID or ϕ) and the standard error of measurement (SEM) as in the following equations:

$$\phi = \frac{\sigma_S^2}{\sigma_S^2 + \sigma_{ABS}^2} \quad (1)$$

$$SEM = \sqrt{\sigma_{ABS}^2} \quad (2)$$

where σ_{ABS}^2 is the absolute measurement error variance. For example (and for the sake of simplicity), in a two-facet random design:

$$\sigma_{ABS}^2 = \frac{\sigma_i^2}{n'_i} + \frac{\sigma_j^2}{n'_j} + \frac{\sigma_{si}^2}{n'_i} + \frac{\sigma_{sj}^2}{n'_j} + \frac{\sigma_{ij}^2}{n'_i n'_j} + \frac{\sigma_{sij}^2}{n'_i n'_j} \quad (3)$$

where n_i and n_j could be, respectively, the number of days (i.e., n_D) and trials n_{Tr} averaged when different D-studies are planned. In the same manner, Equation 3 could be expanded when a three-facet random design is used (as in the case of the present study). It can be seen from equations (1 and 3) that increasing n_i and n_j will effectively improve the reliability (ID increases and SEM decreases). To determine the effect of different measurement strategies to increase the reliability results (ID and SEM), D-studies were simulated using different n_{Tr} (2, 4, 6, 8, 10, 12, 14, 16, 18, 20 trials). Measurement strategies requiring measures performed in different tests or different days were not simulated because this is not very practical. The index of dependability, corresponding to the proportion of variance explained by the Subject factor, is analogous to the frequently used intra-class correlation coefficient or ICC[25]. Like the ICC, the ID ranges between the values of 0 (no reliability) and 1 (perfect reliability), thus, it was interpreted the same way as the ICC: < 0.40 – poor, 0.40-0.75 – moderate and > 0.75 – excellent[8]. The SEM provides an indication of the absolute reliability of the measure (in the same units of measurement). To better judge the relative importance of SEM values may also be expressed as a percentage (%SEM) of the grand mean calculated across days, which is analogous to the coefficient of variation (CV). This choice has the disadvantage of giving large %SEM values when the mean is around zero. However, in such situations, IDs give a better indication of reliability.

4.4 Results

The number of subjects included in the analyses varied depending on the variable and muscle. For the reflex latency variables, the numbers of subjects were: SHEWHART (n=9-15 [NP], n=9-12 [PRE15]), AGLR (n=11-15 [NP, PRE15]), and WAVELET (n= 12-15 [NP], n=9-13 [PRE15]). For the reflex amplitude variables, the number of subjects was: RatioPeak (n=13-15 [NP], n=14-15 [PRE15]) and RatioArea (n=14-15 [NP, PRE15]. For the kinematics variables, all subjects (n=15) were included in the analyses.

For the EMG variables (reflex latencies and amplitudes corresponding to all methods, there were no significant differences ($P > 0.05$) for almost all of the main factors (DAY, TEST, TRIAL) and their interactions. In fact, there were only two cases (out of a possible 210 cases) with a difference in one of the main factors and five other cases where one of the four possible interactions was significant.

Although no statistical analysis was performed because the same subjects were not used in all computerized methods, it appears that there is no difference between muscle groups and between pre-loading conditions for the reflex latencies. However, some trends were observable in the results, with lower latencies at lower electrode sites and in the PRE15 condition (Figure 4.3, Table 4.1). Reflex latencies detected using the SHEWHART method were lower than the two other methods (Figure 4.3). For both conditions, a tendency was observed whereby, the reflex latencies for SHEWART < AGLR < WAVELET. It appears that no significant interaction was obtained. Similarly, reflex amplitudes were consistent between muscles and pre-loading conditions. While trends were observed in the results, these were in different direction for RatioPeak (NP < PRE15) and RatioArea (NP > PRE15).

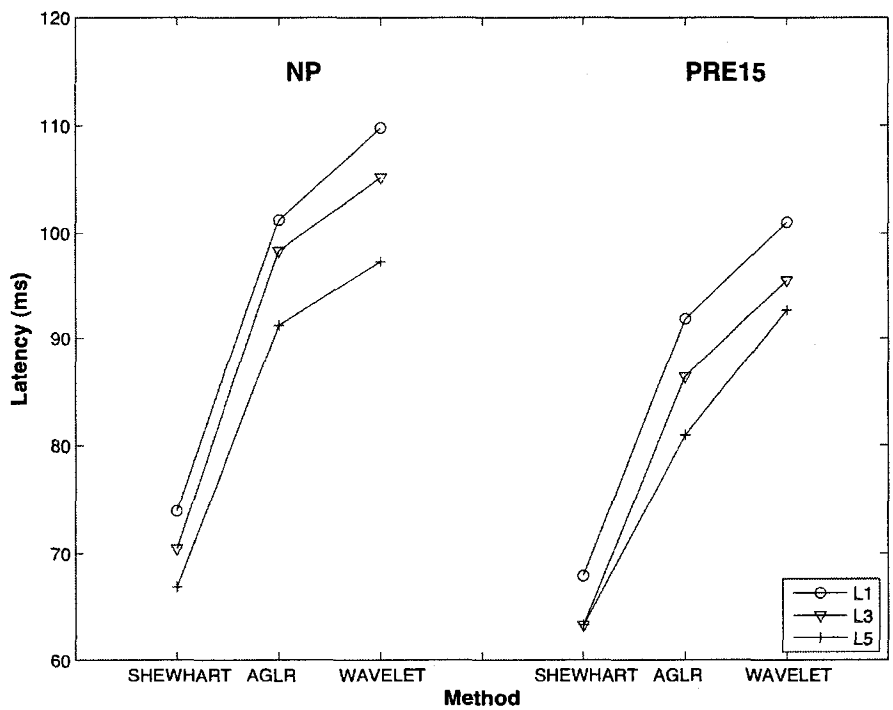

Figure 4.3: Average EMG reflex latencies (ms) for each of the three EMG onset detection methods for the no pre-load (NP) and pre-load (PRE15) conditions

4.4.1 Generalizability Study

4.4.1.1 Electromyography

Results were slightly improved by averaging across the muscle pairs. Therefore, only the results of the homologous muscle pairs will be reported in the subsequent text. The results of individual muscles may be found in Tables 4.1 and 4.2.

EMG reflex latency

For the reflex latency variables, the average (n=3 muscle pairs) of the proportion of variance attributed to the Subject (σ_S^2) in NP was 23.4% (SHEWHART), 40.3% (AGLR), and 31.3% (WAVELET). In PRE15, the corresponding values were lower: 8.1% (SHEWHART), 13.7% (AGLR), and 11.1% (WAVELET). The proportion of variance attributed to the Subject (σ_S^2) corresponds to the ID (or ICC) when n_D, n_{Te} and n_{Tr} are equal to 1. Therefore, the reliability of EMG reflex latency variables was qualified as poor to moderate, depending on the method of onset detection and on the muscle group.

For both NP and PRE15 conditions, the contribution of the day (σ_D^2), test (σ_{Te}^2), and trial (σ_{Tr}^2) facets was less than 5% while for most of the 2-way interactions (σ_{DTe}^2, σ_{DTr}^2, σ_{TeTr}^2), the variance was less than 8% (results not presented in the Table 4.1). On the other hand, the proportions of variance corresponding to the Subject × Day interaction was higher with values reaching up to 14% in the PRE15 condition (Table 4.1).

A large proportion of the measurement variability was attributed to 3-way interactions ($\sigma^2_{SDTe}, \sigma^2_{SDTr}$, σ^2_{STeTr}, σ^2_{DTeTr}), contributing between 0% and 24% and between 0% and 19% in NP and PRE15, respectively (not reported in Table 4.1). However, the largest proportion of variance was attributed to the highest order 4-way interaction (σ^2_{SDTeTr}), containing the residual error. In NP, the average was 37% (SHEWHART), 42% (AGLR), and 35% (WAVELET). Likewise in PRE15, the average was 57% (SHEWHART), 51% (AGLR), and 59% (WAVELET).

EMG reflex amplitude

In NP, the average (n=3 muscle pairs) σ^2_S was 43% (RatioPeak) and 30% (RatioArea) (Table 3.2) and in PRE15 the average σ^2_S was 42% (RatioPeak) and 25% (RatioArea). Overall, when n_D, n_{Te}, and n_{Tr} are equal to 1, the EMG reflex amplitude variables would be qualified having poor to moderate reliability.

For both RatioPeak and RatioArea, the contribution of the day (σ^2_D), test (σ^2_{Te}) and trial (σ^2_{Tr}) facets was less than 7% (NP) and less than 4% (PRE15), indicating very minimal systematic errors (results not presented in Table 3.2). The error variances related to most 2-way interactions (σ^2_{DTe} σ^2_{DTr} σ^2_{TeTr}) were, on average, small (<6%) for both conditions (not presented in Table 4.2). In both NP and PRE15 conditions, the contribution of σ^2_{SD} ranged from 0%-24%, σ^2_{STe} ranged from 0%-11%, and σ^2_{STr} from 0%-9% depending on the variable and muscle (Table 3.2).

A large proportion of the measurement variability was attributed to 3-way interactions ($\sigma^2_{SDTe}, \sigma^2_{SDTr}, \sigma^2_{STeTr}, \sigma^2_{DTeTr}$), contributing between 0% and 24% in both NP and PRE15 (not reported in Table 3.2). In addition, an even larger proportion was attributed to σ^2_{SDTeTr}, the highest order interaction (4-way) and the residual error. In NP, the average σ^2_{SDTeTr} was 22% (RatioPeak) and 32% (RatioArea). In PRE15, the average σ^2_{SDTeTr} was 19% (RatioPeak), and 37% (RatioArea).

4.4.1.2 Kinematics

The contribution of the subject (σ^2_S) facet ranged from 40% (angular velocity) to 45% (displacement) for the NP condition and from 20% (maximal acceleration) to 27% (maximal velocity) for PRE15 (Table 4.3). For both conditions, contribution of the day (σ^2_D), test (σ^2_{Te}) and trial (σ^2_{Tr}) facets was less than 10%. The error variances related to most 2-way interactions (σ^2_{SD}, σ^2_{STe}, σ^2_{STr}, $\sigma^2_{DTe}, \sigma^2_{DTr}, \sigma^2_{TeTr}$) were, on average, small (<4%) for both conditions (those greater than this are shown in Table 4.3). The measurement variability attributed to the 3-way interactions ($\sigma^2_{SDTe}, \sigma^2_{SDTr}$, σ^2_{STeTr}, σ^2_{DTeTr}) was less than 6%. However, a large proportion of the measurement variability was attributed to σ^2_{SDTe}, where the contribution ranged between 23% (maximal acceleration) to 36% (maximal velocity) during NP, and between 7% (average velocity) and 46% (maximal acceleration) during PRE15 (Table 4.3). The highest order interaction, combined with the residual error, also contributed to a large proportion of the variability. The σ^2_{SDTeTr} attributed between 6% (maximal velocity) and 17% (average velocity) during NP, and between 15% (maximal acceleration) and 60% (average velocity) in during PRE15.

Table 4.1: Mean (ms), standard deviation [SD] (ms), standard error of measurement [SEM] (ms) and relative magnitude (%) of the variance components from the G-study for the reflex latency determined using the three different onset-determination methods

	NP								PRE15							
Muscle[a]	Mean	SD	SEM	σ^2_S	σ^2_{SD}	σ^2_{STe}	σ^2_{STr}	σ^2_{SDTeTr}	Mean	SD	SEM	σ^2_S	σ^2_{SD}	σ^2_{STe}	σ^2_{STr}	σ^2_{SDTeTr}
SHEWHART																
1	77	17	19	23	0	0	0	34	70	19	21	13	0	0	0	54
2	73	16	17	8	6	0	3	43	67	14	17	0	3	0	1	46
3	70	15	14	23	4	0	0	54	63	16	21	0	11	16	3	64
4	71	15	19	7	7	15	2	68	64	18	21	13	0	0	6	45
5	68	12	11	35	8	3	0	45	62	16	20	0	0	3	12	65
6	65	13	16	0	0	22	3	66	64	16	23	0	11	9	8	66
1+2	74	13	15	24	0	0	3	15	68	13	15	10	0	0	0	49
3+4	70	13	12	24	7	3	3	56	63	13	14	15	0	0	0	55
5+6	67	10	10	23	2	18	4	41	63	11	16	0	14	3	9	67
AGLR																
1	100	14	14	17	0	0	3	50	91	17	18	16	0	1	6	35
2	102	16	15	38	1	1	2	52	88	16	15	31	0	0	0	46
3	99	15	14	28	2	2	8	51	82	16	20	7	4	0	0	51
4	97	13	12	35	0	0	0	40	86	16	19	8	4	0	0	63
5	92	14	11	38	0	0	0	32	80	16	18	0	3	3	5	66
6	91	12	13	8	7	7	0	61	82	16	19	0	1	0	10	80
1+2	100	12	11	59	0	0	0	41	89	14	13	30	0	2	0	25
3+4	98	11	10	37	0	4	4	44	84	12	14	11	0	0	0	53
5+6	91	1	9	36	0	0	0	41	81	12	16	0	3	0	12	77
WAVELET																
1	110	15	16	11	8	13	7	56	97	15	14	25	0	0	0	47
2	110	15	15	34	5	8	7	40	100	14	15	6	2	0	0	77
3	105	13	16	16	0	0	12	44	93	16	19	16	0	0	3	62
4	104	14	13	30	0	1	6	50	96	17	21	4	12	0	0	58
5	98	12	11	20	12	4	11	38	91	13	14	12	0	0	0	54
6	97	11	11	22	4	3	0	53	93	14	17	6	1	0	9	77
1+2	110	12	12	37	3	9	6	36	97	10	12	17	0	2	0	29
3+4	105	12	11	33	0	0	9	36	93	13	15	9	10	0	0	76
5+6	97	10	9	24	8	7	8	34	93	10	12	7	6	0	4	72

[a] 1: LONG-L1-L, 2: LONG-L1-R, 3: ILIO-L3-L, 4: ILIO-L3-R, 5: MULT-L5-L, 6: MULT-L5-R; 1+2: average of longissimus muscle pairs (L1); 3+4: average of iliocostalis lumborum muscle pairs (L3); 5+6: average of multifidus muscle pairs (L5)

Table 4.2: Mean, standard deviation [SD], standard error of measurement [SEM], and relative magnitude (%) of the variance components from the G-study for the two different methods of determining reflex amplitude

	NP								PRE15							
Muscle[a]	Mean	SD	SEM	σ^2_S	σ^2_{SD}	σ^2_{STe}	σ^2_{STr}	σ^2_{SDTeTr}	Mean	SD	SEM	σ^2_S	σ^2_{SD}	σ^2_{STe}	σ^2_{STr}	σ^2_{SDTeTr}
RatioPeak																
1	5	2	2	33	14	14	0	20	7	3	3	16	26	3	4	35
2	6	3	4	21	0	0	7	39	9	4	4	26	18	0	0	22
3	5	3	3	33	16	10	0	21	6	3	3	33	6	0	0	28
4	5	3	3	34	1	0	3	26	8	4	3	44	4	0	0	17
5	5	3	3	47	1	0	0	36	10	5	4	42	1	0	2	17
6	5	2	3	42	0	0	3	18	11	5	4	43	3	0	0	17
1+2	6	2	2	35	5	4	3	30	8	3	3	32	15	0	0	25
3+4	5	2	2	46	2	1	1	23	7	3	3	45	5	0	0	18
5+6	5	3	2	49	0	0	0	10	11	5	3	50	1	0	0	13
RatioArea																
1	14	11	12	7	16	7	1	61	10	7	8	10	23	0	3	55
2	14	11	12	19	0	9	2	34	8	5	5	23	1	0	0	41
3	15	14	13	32	0	8	0	30	11	11	12	24	0	0	7	41
4	14	13	13	17	9	10	3	48	8	5	5	10	1	12	8	59
5	17	14	13	31	0	0	0	34	7	6	5	35	0	0	0	12
6	16	11	10	28	2	13	0	24	7	4	4	25	0	0	0	35
1+2	14	9	9	23	1	4	0	37	9	5	5	19	24	0	0	45
3+4	15	11	9	38	2	11	0	29	9	7	7	21	1	1	9	49
5+6	17	11	10	31	1	2	0	30	7	4	4	36	0	0	0	17

[a] 1: LONG-L1-L, 2: LONG-L1-R, 3: ILIO-L3-L, 4: ILIO-L3-R, 5: MULT-L5-L, 6: MULT-L5-R; 1+2: average of longissimus muscle pairs (L1); 3+4: average of iliocostalis lumborum muscle pairs (L3); 5+6: average of multifidus muscle pairs (L5)
[b] SD: Standard Deviation

Table 4.3: Mean (units), standard deviation [SD] (units), standard error of measurement [SEM] (units) and relative magnitude (%) of the variance components from the G-study for the kinematic variables

	NP								PRE15							
Variable (units)	Mean	SD	SEM	σ^2_S	σ^2_{SD}	σ^2_{STe}	σ^2_{SDTe}	σ^2_{SDTeTr}	Mean	SD	SEM	σ^2_S	σ^2_{SD}	σ^2_{STe}	σ^2_{SDTe}	σ^2_{SDTeTr}
Angular Displacement (°)	8	2	2	45	3	0	24	15	8	2	2	25	16	0	13	40
Average Angular Velocity (°/s)	20	4	3	40	0	0	30	17	20	3	3	20	2	11	7	60
Maximal Angular Velocity (°/s)	35	6	5	41	0	0	36	6	34	5	4	27	12	9	14	33
Maximal Angular Acceleration (°/s^2)	510	134	100	45	0	9	23	8	571	193	162	20	0	10	46	15

4.4.2 Decision Study

Simulations were done only for EMG variables averaged across muscle pairs because it was obvious, from the G-study results, that this strategy was efficient to increase reliability. As expected, the trend resulting from the simulation studies for all variables (EMG and kinematics) was that as the number of trials averaged increased, so did reliability (i.e., higher index of dependability and lower standard error of measurement), as illustrated in Figure 4.4. However, the increased generally leveled off with the use of six to eight trials.

For the EMG variables, it is difficult, if not impossible, to reach excellent reliability (ICC > 0.75) for almost all of the of the EMG variables. With the exception of one case (i.e., using the AGLR method, only 6 trials are needed for the iliocostalis muscle), excellent reliability cannot be reached with less than 100 trials on one testing day (simulations were stopped at 100 trials) because the equation followed an asymptotic behaviour. However, in rare cases (depending on the variable and muscle) where two or more days are averaged, the target ICC (0.75) could be reached with less than 20 trials. For example, using the AGLR method, the target ICC could be reached by averaging, over two days, four and six trials for the muscle pairs at L1 and L5, respectively. Similarly, the target was reached by averaging, over two days, seven trials for RatioPeak (L3) and 15 trials for RatioArea (L3). Obviously, the number of trials needed to be averaged decreased as the number of days increased. However, in many cases, even if five days of averaging was used, it would still take more than 100 trials on one testing day before the target ICC could be reached.

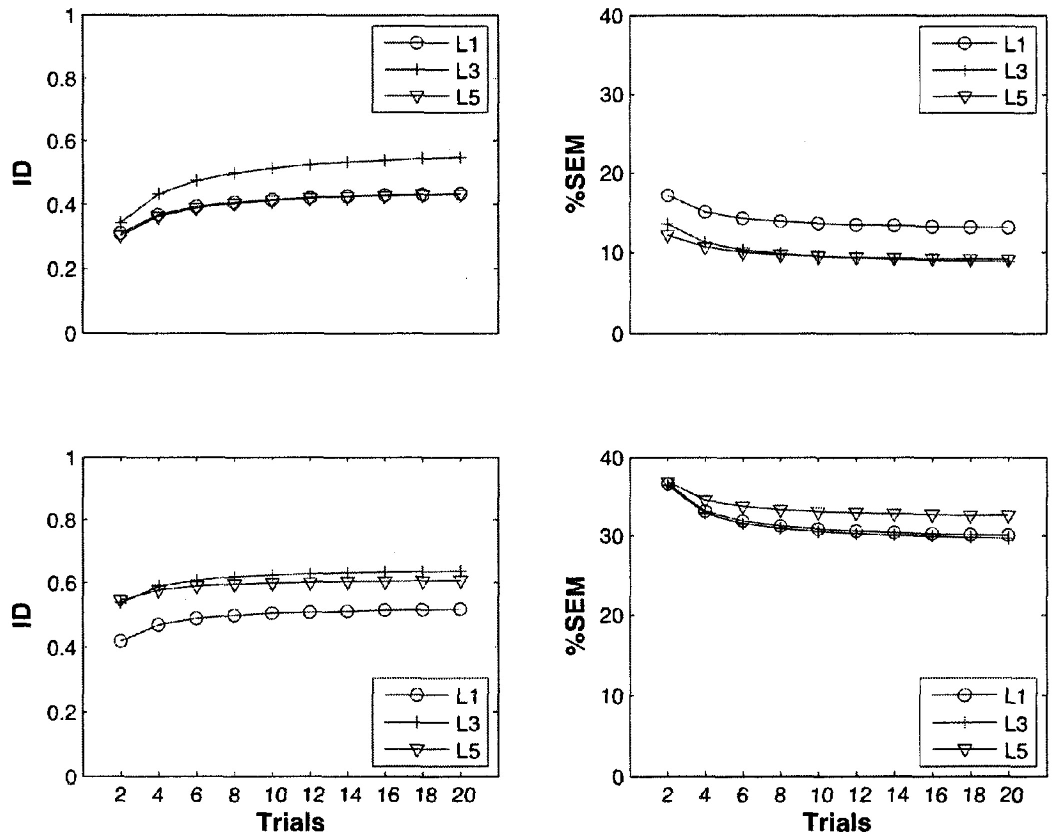

Figure 4.4: Reliability statistics (index of dependability [ID] and standard error of measurement [SEM]) as a function of the number of trial (2, 4, ..., 20) averaged over one day obtained from the decision (D-) study. The example used here is during the NP condition for the EMG reflex latency (SHEWHART method) and EMG reflex amplitude (RatioPeak) for all muscle pairs.

4.5 Discussion

The current results may apply only to a sudden load paradigm where the load is applied directly on the trunk and where the lower limbs and pelvis are well stabilized. Whether different reflex responses could be obtained using different combinations of experimental conditions is unknown, though previous results suggest that this might be the case[16].

The main purpose of this present investigation was to assess the reliability of EMG reflex latency, EMG reflex amplitude and trunk kinematics variables, during a sudden loading perturbation, using the generalizability theory[24] as a framework. Unlike the classical test theory, the generalizability analysis allows both researchers and users to estimate both the magnitude and relative contribution of different sources of measurement error. Being aware of these errors would allow for the investigators to correct for them. In addition, investigators would be able to make decisions as to different measurement strategies that could provide them with optimal reliability depending on their resources.

4.5.1 Generalizability Study

The proportion of variance attributed to the subject, corresponding to the ICC when n_D, n_{Te} and n_{Tr} are equal to 1, showed that the reliability of the EMG and kinematics variables to describe the reflex response during a sudden loading perturbation ranged from poor to moderate. Recent evidence shows that the reflex response is largely accounted for by the movement velocity of the trunk[20]. Even though relatively large differences in angular kinematics must be present to affect muscle reflex responses, as discussed elsewhere[1], a good control of this confounding variable may help reducing the inter-subject variability of reflex responses (increase diagnosis capability) though this remains to be substantiated. The corresponding ICCs were low (Table 4.3, σ_S^2 values) but this might be explained by the relatively low inter-subject variability in the trunk kinematics generated by our method to adjust the pre-load and sudden-load according to the inertial properties of the trunk. In contrast, the %SEM values were less than 15% for the angular velocity variables. However, such relatively low variability in the perturbation kinematics was apparently not sufficient to generate reliable reflex responses.

Surprisingly, having a preload before the application of the perturbation was not more reliable than not having a preload. This was more so for the EMG reflex latency and kinematics variables. It would have been expected that the preload condition would result in higher reliability values since this condition was meant to help pre-activate the back muscles in a standardized manner from one trial to another. However, these results are maybe in line with Stokes et al. (2006) who found that muscle responses to perturbations were detected more frequently in the low preload condition than in the high preload condition.

Averaging measures across homologous muscles increased the reliability of the EMG variables (even though reliability remained either poor or moderate). In few cases was the ICC better than both muscles individually but in all cases the SEM decreased.

By averaging across homologous muscles, we can eliminate any chance of choosing a unilateral muscle, which would have poorer reliability. This procedure of averaging across bilateral back muscles has previously been shown to improve reliability with other EMG indices[15]. This measurement strategy, that cannot be applied to muscle groups of lower and upper limbs, has the advantage to be easy to apply.

The presence of any systematic errors should first be verified in a reliability study. In the present investigation, the variances explained by the day (σ_D^2), test (σ_{Te}^2), and trial (σ_{Tr}^2) facets were minimal (< 6%), with a few exceptions not exceeding 10%. The low σ_D^2 and σ_{Tr}^2 and σ_{Te}^2 variance values suggest that learning was minimized, while the low σ_{Tr}^2 and σ_{Te}^2 variance values would further indicate that muscular fatigue did not build-up across trials. These were further substantiated by the ANOVA results, where there were practically no significant differences (n = 2 cases as outlined above) for the Day, Test, and Trial factors.

The starting "neutral" posture of the trunk before the sudden loading, which could affect the initial stretch of the spinal muscles and hence the reflex response, was standardized across trials, tests and days. However, this posture may slightly vary in different directions between subjects, which could affect all interaction terms involving the subject variance. Likewise, the placement of EMG electrodes between days was standardized using a template repositioned on the subject anatomical landmarks. However, the relative position of each electrode relative to the targeted muscle may again slightly fluctuate in different directions between subjects and days, which could affect the interaction terms involving the subject and day variances. Finally, repositioning of the subject in the apparatus between tests and days would be reflected in σ^2_{SD}, σ^2_{STe}, σ^2_{DTe} and σ^2_{SDTe} in both kinematics and EMG variables. Unfortunately, although we have identified potential sources of variability that may contribute to the interaction terms (trunk posture, electrode positioning, subject positioning in the apparatus), it is difficult to do better.

A large contribution to the variability of the measurement was the highest order interaction σ^2_{SDTeTr}, which contains the unexplained random variance (i.e., stochastic nature of the EMG signal) and error variance attributed to facets not identified in the study (i.e., trunk muscles state, excitability of the spinal motoneuron pool). For example, the co-contraction level of the abdominals was attempted to be minimized with the use of biofeedback but the level of co-contraction is more or less difficult to control in different individuals. This co-contraction of the trunk muscles serves to increase spinal stability[4,10], which would increase muscular activation and thus result in smaller muscular responses to a perturbation[14,30]. Further, the preactivation level of all trunk muscles should be comparable between subjects. Increasing the preactivation levels of trunk muscles increases spinal stiffness and consequently decreases trunk excursion, thereby decreasing the frequency and amplitude of the reflex responses[36]. The only way to decrease the effect of random errors is to average the score of several trials, which was investigated in the D-study. Perhaps using a different protocol such as that used by Moorhouse and Granata (2007), this variability in the results could be minimized.

4.5.2 Decision Study

In general, it was difficult, if not impossible, to reach excellent reliability (ICC > 0.75) for almost all of the of the EMG variables. In the majority of the cases, more than 100 trials are needed to achieve excellent reliability. This is in no way practical and not feasible without fatiguing the subject. Simulations were performed by increasing the number of trials averaged over one day. It would have also been possible to perform simulations by averaging trials across days, which, in a clinical context, would not be very practical. As a cautionary note, although the results demonstrated minimal systematic error due to the Trial factor, it would be very difficult to speculate on the effect of learning or fatigue if more than five trials are performed. The reader is reminded that the present reliability study is based on 5 trials in a row only. Therefore, this must be taken into account when making decisions about the study design.

The ICC values in this study were low (poor to moderate) due to the low inter-subject variability relative to the other sources of variability. In other words, the diagnosis value of back muscle reflex responses is not good. This could have been expected since variability among the subjects is reduced by adjusting the load as a function of trunk inertial properties. However, it should be noted that the SEM is also of importance when evaluating an individual subject/patient. If repeated measures are made and the measurement of changes (e.g. resulting from a treatment) is the priority, then

SEM is the index of interest. The SEM for the reflex latency variables were generally low (<20%), however were slightly higher (<56%) with the reflex amplitude variables. These SEM values decrease as the number of trials increase. The SEM could also help us to see whether adjusting the load was efficient to reduce within-subject variability in trunk kinematics. ICC and SEM values for all variables, muscles, and for both conditions were plotted similarly to those in Figure 4.4. After visual inspection of these plots, it was observed for the majority of the cases that the values levelled-off after six to eight trials. Based on these observations and if we evaluate the cost/benefit of performing several repeated measures, it would be reasonable to suggest that at least eight trials be averaged in order to achieve the best reliability possible. Depending on the tolerance of the study population to the sudden loading paradigm, the researcher may want to weigh the benefits of performing additional trials. Additionally, as mentioned earlier, the effects of learning and/or fatigue are unknown when more trials are performed.

At this point in the discussion, it is important to outline the limitations of the current study. It should be noted that even though 15 participants completed the study, the number of participants included in the analyses for the EMG indices varied depending on the method and muscle. This, in effect, made the groups of participants different between methods and muscles. Lastly, the results from this study may not be generalized to female participants and to back pain subjects, as they demonstrate different back muscle composition[18,34].

4.6 Conclusion

The results of this study show that the reflex response of the back muscles is inherently variable using this sudden loading protocol, which can possibly be attributed to the difficulty in standardizing back muscle elongation (amplitude, velocity) across participants, days and trials. Both EMG and kinematics variables yielded poor to moderate reliability. Based on these findings and the fact that reliability reached a plateau after averaging 6 to 8 trials, it would be reasonable to suggest a minimum of eight trials be averaged during the same testing day to achieve acceptable reliability. Furthermore, improved reliability could be achieved by averaging the scores of bilateral muscles.

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CHAPTER 5

A LABORATORY STUDY TO QUANTIFY THE BIOMECHANICAL RESPONSES TO WHOLE-BODY VIBRATION: THE INFLUENCE ON BALANCE, REFLEX RESPONSE, MUSCULAR ACTIVITY AND FATIGUE

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5.1 Abstract

To determine the acute effects of WBV on the sensorimotor system and potentially on the stability of the spine, different biomechanical responses were tested before and after 60 minutes of sitting, with and without vertical whole-body vibration (WBV), on four different days. Postures adopted while sitting and the simulated WBV exposure corresponded to large mining load haul dump (LHD) vehicles as measured in the field. Twelve males performed trials of standing balance on a force plate and a sudden loading perturbation test to assess back muscle reflex response, using surface electromyography (EMG). This latter test also allowed to assess if any muscle fatigue occurred as a result of the exposure. First of all, it was shown that back muscle activity while sitting with vibration was significantly higher as compared to back muscle activity while sitting with no vibration. However, WBV per se elicited very few effects on the outcome variables and thus not supporting our hypothesis that WBV had any effect on spinal stability. Though WBV may not have elicited any effects, new findings have emerged concerning the effect of sitting on muscle fatigue and balance. It was shown that sustaining trunk sitting postures corresponding to mining vehicle operators generates back muscle fatigue. Unexpectedly, standing balance was also improved. The possible explanations and relevance of these findings are discussed.

Relevance to industry

Occupational groups exposed to WBV while sitting are at increased risk for low back disorders. The results of this study do not support the possible injury pathway linking

WBV and back pain via sensorimotor deficits. Unexpectedly, it appears that sitting per se may affect the sensorimotor system but this may only apply to sitting postures corresponding to driving mining vehicles.

Keywords: whole-body vibration; sitting; electromyography; balance, reflex response, muscle fatigue, back muscles, low back pain

5.2 Introduction

A large body of evidence supports the importance of physical factors in the development of LBD (Nachemson and Jonsson, 2000). Among such physical exposures encountered in working conditions, whole body vibration (WBV) has repeatedly been identified as a risk factor for low back pain (Bernard, 1997; Bovenzi, 1996; Lis et al., 2007; Seidel, 1993). In Canada, the United States and some European countries, an estimated 4 to 7% of employees are exposed to potentially harmful levels of WBV (Bernard, 1997).

Several authors have concluded that there is an association between LBD and WBV (Bovenzi and Hulshof, 1998; Lings and Leboeuf-Yde, 2000), however, the injury mechanisms linking WBV to low back disorders needs to be better understood. The difficulty is that, in occupations in which workers are exposed to WBV, other physical risk factors are also present such as awkward postures, prolonged sitting (Lis et al., 2007) as well as loading and unloading materials from a vehicle.

Evidence arising from *in vitro* studies has shown that prolonged vibration exposure may cause spine pathology (and nociception) through mechanical damage, most notably to the vertebrae, vertebral endplates, intervertebral discs, and low back musculature (Wikström et al., 1994). According to Panjabi's lumbar-stability hypothesis (Panjabi, 1992), any impairment in the passive (discs, ligaments, vertebrae), active (muscles) and/or neural subsystems of the spine may in fact lead to lumbar spine instability and possibly to intervertebral buckling and its associated tissue damage. An increasing body of scientific evidence now supports this hypothesis (Preuss and Fung, 2005). However, using this hypothesis as a conceptual framework implies that many injury pathways may be involved.

The muscles, the active subsystem, are influenced by vibration. Mechanical vibration (namely between 30 and 120 Hz) directly applied to a muscle belly or tendon elicits the tonic vibration reflex (TVR) (Desmedt, 1983; Vermeersch et al., 1986), a neuromuscular response caused by excitation of muscle spindles leading to enhanced muscle activity. The TVR has been suggested to occur in the back muscles at frequencies between 1 and 5 Hz (Seidel, 1988), however, the evidence to support this phenomenon happening at lower frequencies is still not conclusive. This increased muscle activity is necessary to dampen the vibratory waves (Wakeling and Nigg, 2001), though could lead to muscle fatigue, which has been shown to effect neuromuscular coordination (Ng et al., 2003) and proprioception (Taimela et al., 1999), thus leaving the spine at increased risk of injury. Few laboratory studies have demonstrated back muscle fatigue after exposure to seated WBV (Hansson et al., 1991; Wilder et al., 1994). The shortcomings of these studies are the short exposure duration (i.e., < 10 minutes) and different conditions performed on the same day (Wilder et al., 1994), which makes it difficult to separate the effects of a single condition.

WBV may also impair some neuro-sensory functions such as back muscle reflex responses and postural balance. Delayed trunk muscle reflex responses (Cholewicki et al., 2005) as well as poor postural balance (Takala and Viikari-Juntura, 2000) could increase the risk of low back injuries. To the authors' knowledge, only one published study has investigated the effect of WBV on back muscle reflex response (Wilder et al., 1996), which reports an increase in the latency and the magnitude of the response of the erector spinae muscles after a 40-min exposure to vertical vibration.

Studies examining the effects of WBV on postural balance are also limited (Martin et al., 1980; McKay, 1972; Seidel et al., 1980). Exposure to WBV has resulted in increases (though not always significant) in postural sway at fixed sinusoidal frequencies ranging from 12.5 Hz – 18 Hz (Martin et al., 1980; McKay, 1972). The external validity (i.e., in relation to the workplace conditions experienced by drivers) of these studies could be questioned with this use of only pure sinusoidal WBV. This issue was addressed in a study that exposed subjects for 40 min to vertical WBV with a frequency spectrum resembling a shuttle car operation (Cornelius et al., 1994). They found no differences before and after WBV on any of the balance measures. However, only one trial was performed, thus limiting the reliability and sensitivity of this measure.

According to the various effects of WBV on the three subsystems, the lumbar spine could be at greater risk for injury, especially if the residual deficits take time to recover after exposure. For example, a worker who is first exposed to a period of WBV and then immediately following, must perform other activities (descending from their vehicle, manual handling tasks) may be at greater risk for low back injury if lumbar stability cannot be assured due to an affected neuro-sensory subsystem.

Evaluation of the biomechanical responses to seated WBV exposure is important in improving our understanding of the role WBV might play in the development of LBP or injury. Therefore, the purpose of the present study was to evaluate the acute effects of seated WBV exposure on the sensorimotor system (balance, back muscle activity, back muscle fatigue and back muscle stretch reflex response). It attempted to improve on the methodological weaknesses of previous studies investigating biomechanical effects due to seated WBV by increasing the duration of exposure and by including a control (no vibration) condition. Additionally, real-life working conditions (i.e., vibration exposure, postures) experienced by operators of large underground mining vehicles were "reproduced" in a laboratory setting to make the simulation as realistic as possible.

5.3 Methods

5.3.1 Subjects

Twelve healthy males were recruited to participate. Participant mean ± SD age, height and weight were 22 ± 2 years, 1.8 ± 0.1 m, and 77 ± 9 kg, respectively. The exclusion criteria were: presence of a systemic or degenerative disease, neurological disease, musculo-skeletal problem, low-back pain requiring medical treatment in the last 12 months, a positive response to the Baecke Habitual Physical Activity Questionnaire (Baecke et al., 1982). Furthermore, to facilitate surface EMG measures, anyone was excluded if their body mass index exceeded 30 kg/m^2. Before their participation, all subjects were informed of the experimental protocol and of its potential risks, and signed a consent approved by the university Research Ethics Committee.

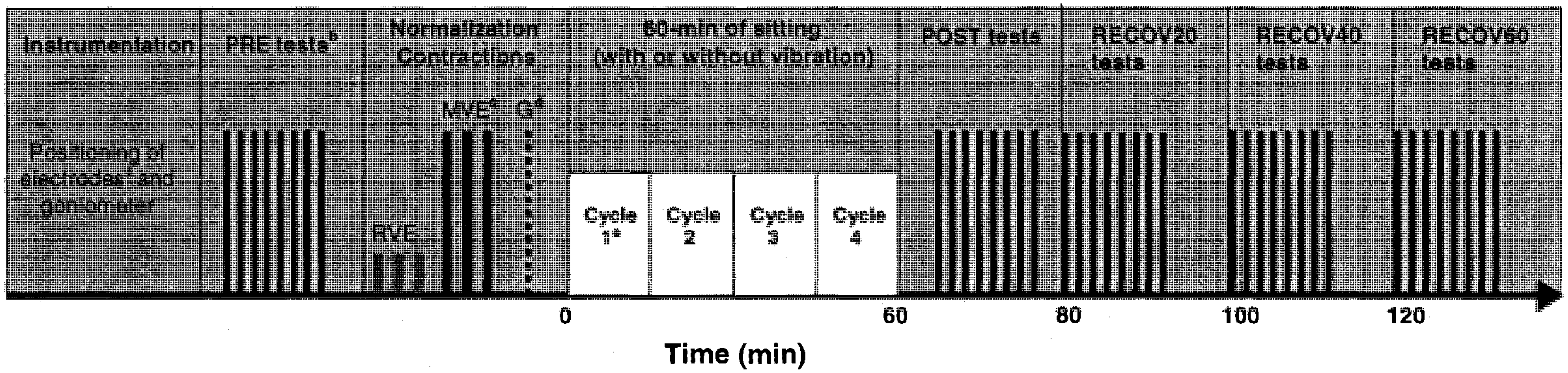
5.3.2 Study design, tasks and general procedures

5.3.2.1 Study design

A repeated-measures design was used to study the effect of a 60-min exposure (vibration/no-vibration) on four biomechanical measures using two different tests (balance/ sudden loading). Participants performed a total of four experimental conditions (2 tests [balance/sudden loading] × 2 exposures [vibration/no vibration]), each on a separate day, with at least one day and no more than one week between testing sessions. The conditions were presented to the subjects according to a counterbalanced design. For each test, measurements were taken before (PRE) and immediately after (POST) the 60 minutes of exposure. To evaluate if there were any possible effects that remained after the exposure (recovery), additional measurements were taken at 20 (RECOV20), 40 (RECOV40), and 60 (RECOV60) minutes after the end of exposure.

5.3.2.2 Procedures

A detailed illustration of the measurement protocol is presented in Figure 5.1. The maximum duration of the total testing protocol was 3 h 15 min (when the sudden loading test was performed).



[a] Number of electrodes positioned varied depending on the measurement technique (Balance: n=2; Sudden loading: n=8).
[b] Each bar represents one trial (n=8) for both balance and sudden loading test. The time to perform the eight trials was a maximum of 15 minutes.
[c] Reference (RVE) and maximal (MVE) Voluntary Electrical activity obtained during submaximal and maximal contractions, respectively. Performed only on days when the sudden loading tests were executed.
[d] Reference posture for the goniometer.
[e] Please refer to Table 1 for a complete description of the postures adopted during each cycle

Figure 5.1: Time line of a typical experimental session (time period not to scale).

Reference contractions

To normalize future EMG signals, two types of reference contractions were performed for the back muscles: sub-maximal and maximal voluntary contractions (MVC). MVCs were performed only during the sessions in which the sudden loading test was performed.

Sub-maximal voluntary contractions. Subjects performed three 5-s sub-maximal contractions to obtain a reference voluntary electrical activity (RVE). The subject sat on a rigid seat, with his hips and knees flexed at approximately 90°, and flexed his shoulders with his arms straight in front until they were approximately parallel to the floor, with 1-kg weights in each hand.

Maximal voluntary contractions. Three 5-s maximal contractions were performed, while sitting on the vibration simulator with the pelvis stabilized (using a strap), to obtain a maximal voluntary electrical activity (MVE). A strain-gauge type dynamometer was fixed horizontally onto the simulator with a chain attached to a harness fastened around the subject's chest. The subject was instructed to gradually pull against the chain to solicit the back muscles. Strong verbal encouragement as well as visual feedback of the strain-gauge signal was provided to obtain maximal contractions. Two minutes of rest was allocated between each trial.

Postural balance test

Postural sway (or the movement of the centre-of-pressure), evaluated on a force plate, was used as an outcome measure of balance (Winter, 1995). Each participant stood barefoot with both feet parallel on each side of a 2-inch (5.1 cm) T-shaped separator (to ensure that the positioning of the feet was consistent from trial to trial) placed on the surface of the force plate. This separator was always placed at the same position on the force plate and then removed once the subject's feet were positioned. The participant was then instructed to stand quietly with their arms hanging to their sides, their head in a forward-facing position and eyes closed. Eight 60-s trials were taken; this having been previously determined as the number of trials needed to be averaged to yield reliable results (Santos et al., 2006b). Between each trial, the subject stepped off the force plate, and then immediately stepped back on to perform another trial.

Sudden loading test

The reflex response of the back muscles was measured using a sudden loading apparatus (Krajcarski et al., 1999) designed to give an anteriorly directed perturbation of the trunk (Figure 5.2). The zero position for the trunk was when the subject's trunk was upright and still. Before each trial, the subject was asked to position his trunk in the same position, using a visual feedback from the potentiometer, in order to always find the initial reference (zero) position. To pre-activate the back muscles in a standardized manner from trial to trial, a pre-load, equivalent to 15% of the trunk mass (or 48.5% of the total body mass) (de Leva, 1996) was used. To minimize the pre-activation of the abdominal muscles, visual feedback of the two abdominal muscles was displayed for the subject. The load (pre-load and added load) was hidden by a metal screen to remove any visual clues so the effect of anticipation could be eliminated. The added load was released randomly between 5 and 15 seconds after the program had been activated. This added load, equivalent to 35% of the trunk mass, was large enough to minimally solicit the back muscles though not large enough to inhibit the reflex response. Lastly, subjects were instructed to stop their trunk displacement as quickly as possible, once the added load had been released, but to avoid overreacting by extending the trunk backward. Again, eight trials were performed to obtain acceptable reliability (Santos et al., 2006a), with approximately 30 s between trials.

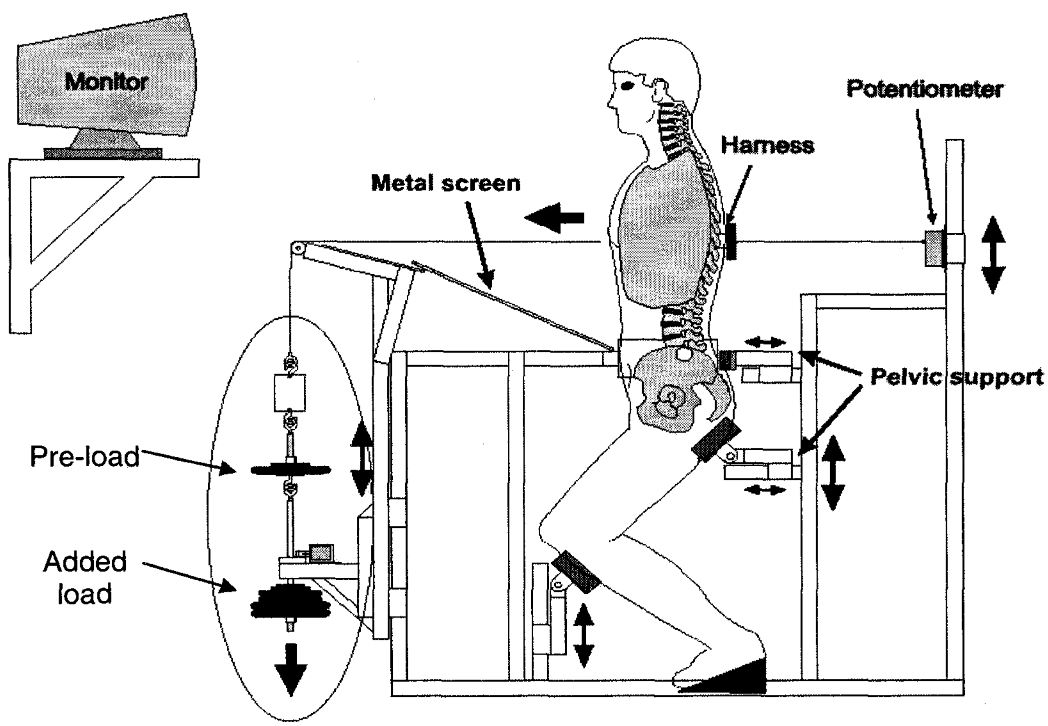


Figure 5.2: Sudden loading apparatus. This apparatus allows for the stabilization of the subject's lower extremities and pelvis. The sudden load was applied via a cable connected to a load cell and attached (at the level of T8) to a harness adjusted on the subject's chest and shoulders. This load, initially held by an electro-magnetic release mechanism, was released from a minimal height of approximately 1 cm to minimize ballistic loading effects, thus assuring the safety of the test.

<u>Seated exposure</u>

During the 60 min of seated exposure (to either vertical random WBV or no vibration), participants were seated on a suspension-type seat with a weight adjustment dial (KAB 525 model, KAB Seating, Vonore, Tennessee, USA) that was mounted on a vehicular vibration simulator. This seat is typically used in large mining load haul dump (LHD) vehicles. The vibration exposure characteristics, to which the subjects were exposed, represented the average spectral signature of large capacity mining vehicles. This spectral signature was obtained from accelerometer signals collected directly in the mines (Boileau et al., 2006; Eger et al., 2005); it ranged between 0.5 and 20 Hz, with the peak frequency centred around 2.7 Hz, and a frequency-weighted average acceleration ($a_{w(0.5\text{-}20Hz)}$) of 0.86 m/s^2.

During the 60-min exposure, the subject placed his hands on the steering wheel (as if he was driving) with the elbows at approximately 90° resting on the armrests. In addition, he adopted postures of an LHD operator during underground mining operations, as identified (Eger et al., 2006) during field observations. It was observed that the trunk of the LHD operators was always inclined approximately 15° forward from the vertical, not using the back rest. Therefore, to prevent the subject from leaning against the backrest, a buzzer was placed behind some padding on the backrest to warn the subject. To simulate the postures adopted by LHD operators, targets (lights) were placed at different locations in the laboratory. When a target lit up, the participant was asked to look at the target; this simulated the viewing location an operator would choose when performing different activities. One cycle, a duration of 15 minutes, consisted of "performing" a combination activities being: 1) forward tramming; 2) mucking (bucket loading); and 3) backward tramming. This cycle (Table 5.1) and was repeated four times.

Table 5.1: Description of the simulated task, postures adopted, and duration in each posture during one cycle of seated exposure

Simulated Task	Subject Posture*	Target Location	Duration (min)
Forward tramming to the muck zone (empty bucket)	• A minimum of 45° of neck rotation (left) • A minimum of 15° of trunk flexion	Target is placed 3 m to the front left corner	5
Mucking	Dynamic posture movements looking left, forward and right	Participants spot a target location which changes from the left to front to right	0.5
Backward tramming (to exit mucking point)	• A minimum of 45° of neck rotation (right) • A minimum of 15° of trunk flexion	Target is placed 3 m to the back left corner	2
Forward tramming full bucket	• A minimum of 60° of neck rotation (left) • A minimum of 15° of trunk flexion	Target is placed 3 m to the front left corner	5
Dumping	Dynamic posture movements looking left, forward and right	Participants spot a target location which changes from the left to front to right	0.5
Backward tramming (to exit the ore pass)	• A minimum of 45° of neck rotation (right) • A minimum of 15° of trunk flexion	Target is placed 3 meters to the back left corner	2

* The targets were positioned laterally so that the described neck postures (rotations) were reached. The 15° trunk flexion was evaluated visually by the investigator throughout the 60-min exposure.

5.3.3. Measurement techniques

All signals were collected, converted to digital signal via a 16-bit A/D converter, and stored on a hard disk for later analysis.

5.3.3.1 Electromyography

Differential pre-amplified (gain: 1000, band-pass filter: 20 – 450 Hz) active surface electrodes (Model DE-2.3, DelSys Inc., Wellesley, MA) composed of two silver bars (spaced 10 mm, 1 mm wide) were used to collect EMG signals during the 60-min exposure (sitting with or without WBV) and during the sudden loading test.

After the skin at the electrode site were shaved, gently abraded and cleaned with alcohol, the electrodes were positioned bilaterally on the longissimus at the level of L1 (LONG-L1-L and LONG-L1-R), iliocostalis lumborum at L3 (ILIO-L3-L and ILIO-L3-R), and multifidus at L5 (MULT-L5-L and MULT-L5-R) following the recommendations of (De Foa et al., 1989). To ensure the same placement of the EMG surface electrodes for the back muscles from session to session, a template using visible anatomical landmarks was used. Additional electrodes were positioned on the right rectus abdominus and right external obliques (following the recommendations of (McGill, 1991). A reference snap-on type surface electrode (Medi-Trace model, Graphic Controls Canada Limited, Gananoque, Ontario, Canada) was positioned on the spinous process at the C7 level.

During the two sessions where postural balance was assessed, only two muscles were investigated (LONG-L1-L and LONG-L1-R) to quantify the muscle responses during the 60-min sitting conditions (with and without WBV).

5.3.3.2 Force plate

Force plate outputs (forces, moments) were collected at a sampling frequency of 100 Hz by a 90 cm x 90 cm force plate (BP900900, Advanced Mechanical Technology, Inc., Watertown, MA).

5.3.3.3 Goniometry

Changes in the posture of the lumbar back were recorded using a bi-axial back goniometer (SG-150, Biometrics, Gwent, Great Britain). Only flexion and extension angles were recorded. The endblock was first attached to the sacral region at S1. With the subject standing upright and the goniometer near minimum length, the telescopic endblock was attached to the back at approximately T12-L1 (depending upon the height of the subject).

5.3.4 Data processing

All data processing and data reduction were performed using MATLAB (Version 7.0, The Mathworks, Natick, MA). The following sections describe how the different outcomes were computed.

5.3.4.1 Outcome measures assessed with electromyography

Post-processing of all EMG signals involved several steps. A notch filter was first applied to eliminate possible 60-Hz electrical noise interference (and its harmonics up to 420 Hz). Then, different processing was applied depending on the outcome to be assessed (muscular activity, reflex responses, muscular fatigue) as described in the next sections.

Back muscles activity

A high-pass digital filter (30 Hz) was used to exclude the electrocardiographic signal (Redfern et al., 1993) and to reduce the potential influence of skin movement on the signal (Hansson et al., 2000). Then, root mean square (RMS) values of back muscle EMG signals were calculated over consecutive time-windows (0.125 s) for each reference contraction and during the 60-min sitting tasks. For each back muscle, the EMG reference values to normalize EMG amplitude variables were obtained by averaging the RMS values of the three submaximal reference contractions (EMG_{RVE}) or by calculating the highest RMS values of the three maximal reference contractions (EMG_{MVE}). Then the signals collected during the 60-min sitting tasks were normalized to either EMG_{RVE} (all four sessions) or EMG_{MVE} (sessions where back muscle reflex was investigated).

To quantify the magnitude of muscle activity during the 60-min sitting tasks, the amplitude probability distribution function (APDF) (Jonsson, 1978) of the normalized EMG RMS values was determined. The 10^{th}, 50^{th} and 90^{th} percentiles (%ile) of the

APDF normalized to MVE ($APDF_{MVE}$) and RVE ($APDF_{RVE}$) were determined for each 15-min posture cycle (i.e., 0-15 min, 16-30 min, 31-45 min, 46-60 min).

Back muscles reflex responses

Signals were filtered using a wavelet method to remove ECG artefact (Blüthner et al., 2002) and then notch filtered to eliminate possible 60-Hz electrical noise and its harmonics (up to 8 harmonics). The EMG signals were first low-pass filtered (2nd order Butterworth filter, 25 Hz cut-off frequency). The reflex responses were determined by two variables: reflex latency and reflex amplitude. The reflex latency was defined as the time from force perturbation to the onset of the EMG response. EMG onset was determined automatically using a 25 ms sliding window and a criteria of two standard deviations above the baseline mean as the threshold (Hodges and Bui, 1996). EMG reflex latencies <30 ms and >150 ms were assumed to be non-reflexive and eliminated from the analyses. To quantify the reflex amplitude, an EMG amplitude ratio was computed as the ratio of the first EMG onset to the average EMG signal during the 250 ms rest period prior to the sudden loading.

For both the EMG reflex latency and amplitude, it has previously been determined that left and right back muscles were not significantly different and that averaging across homologous pairs yields more reliable results (Santos et al., 2006a). Therefore, only the average of the homologous pairs will be presented in the results.

Muscular fatigue

The EMG signal recorded prior to the sudden loading was used to assess any presence of fatigue as there was a constant load applied across trials. The first five seconds before the load was dropped was used to calculate the instantaneous mean power frequency (IMNF) (Karlsson and Gerdle, 2001). MPF values >300 Hz were eliminated; values above this arbitrary cut-off are considered out of the physiological range (Clancy et al., 2005) and are likely explained by a low signal to noise ratio (low contraction intensity). Therefore, data from two subjects were eliminated from the analysis for LONG-L1-R and ILIO-L3-L.

5.3.4.2 Postural sway (balance test)

The COP antero-posterior (COP_{AP}) and medio-lateral (COP_{ML}) coordinates were computed using the force plate outputs (forces, moments) using an in-house C++ program.

The COP time series signals were filtered using a second-order zero phase Butterworth low-pass digital filter with a cut-off frequency of 10 Hz and an in-house program was used to compute a total of 36 summary measures (Prieto et al., 1996). However, only summary measures frequently reported in the literature are presented here (Table 5.2).

Table 5.2: List of abbreviations (alphabetical order) used to describe the COP summary measures

COP summary measure	Description (units)
Area_CE	95% confidence ellipse area (mm^2)
Area_SW	Sway area (mm^2/s)
MPF*	50% power frequency or Median power frequency (Hz)
MFREQ*	Mean frequency (Hz)
MVELO*	Mean velocity (mm/s)

* These measures are computed based on the resultant distance (RD) time series (i.e., the vector distance from the mean COP to each pair of points in the AP and ML time series). Thus, each summary measure also has its AP and ML counterparts that were computed.

5.3.4.3 Goniometry

The goniometer signal was filtered using a second-order zero phase Butterworth filter, with 20 padding points added and an optimal cut-off frequency used. Cut-off frequencies ranged between 0.3-1.3 Hz (no vibration condition) and 0.5-3.7 Hz (vibration condition).

5.3.5 Statistical analyses

Data were analyzed using the NCSS software (Number Cruncher Statistical Systems, Kaysville, Utah, USA). Some of the variables were not normally distributed and thus were corrected using a logarithmic transformation before any statistical analyses were performed. Before averaging some outcome measures (COP parameters, reflex latency and amplitude, IMNF) across the trials (n = 8) to increase their reliability, a one-way ANOVA across the PRE trials, and also across the POST trials, was performed. There were no significant differences across trials for any of the outcome variable and thus, these trials were averaged (n = 8) at each measurement period (PRE, POST, RECOV20, RECOV40, RECOV60). These mean scores were used for subsequent analyses.

To assess whether the conditions of exposure, in terms of body posture as measured by the goniometer, were comparable between the four testing sessions a two-way ANOVA (4 Sessions × 4 15-min posture cycles) with repeated measures was performed on the mean lumbar back angles. In this case, non-significance of the Session factor and of the interaction shows that the conditions are comparable regarding trunk posture. To assess whether exposure to vibration performed on different days resulted in similar EMG activity, a two-way ANOVA (2 Tests × 4 15-min posture cycles) was performed for $APDF_{RVE}$ values. Likewise, exposure to sitting (i.e., no vibration) performed on different days was also assessed to see if this resulted in similar EMG activity.

A two-way ANOVA (2 Vibration conditions × 5 Measurement periods) with repeated measures on both factors was performed to assess the differences between the vibration conditions (vibration, no vibration) and the measurement periods (PRE, POST, RECOV20, RECOV40, RECOV60) and its interaction for the COP summary measures, reflex latency and amplitude, and IMNF.

A two-way ANOVA (2 vibration conditions × four 15-min posture cycles) with repeated measures on both factors was performed on the 10^{th}, 50^{th} and 90^{th} percentile of

$APDF_{MVE}$. An α level of $P \leq 0.05$ was considered statistically significant for all statistical tests. Subsequent *post hoc* tests were performed using Bonferroni multiple comparisons.

5.4 Results

5.4.1 Post-test control measures

The average lumbar posture over the 60 min exposure was 5.0° (SD: 2.8°) and 4.9° (SD: 3.2°) of flexion, for the Vibration and No Vibration conditions, respectively. For the four posture cycles, average angles for the 15-min blocks ranged from 4.6°-5.4° and 8.3°-9.6° for the Vibration and No Vibration conditions, respectively. Results from the ANOVA revealed that there were no significant main effects (*Session* effect: p=0.901; *Time* effect: p=0.618; *Session* × *Time*: p=0.300). Results from the ANOVAs for the 10^{th}, 50^{th} and 90^{th} %ile $APDF_{RVE}$ revealed that there were neither significant Test effects nor any *Test* × *Condition* interactions for any of the muscles.

5.4.2 Muscular activity

Significant main effects for *Condition* were found for some of the muscles (Table 5.3: LONG-L1 on both sides; ILIO-L3-L). Only muscles with significant *Condition* main effects are reported in Table 3. The EMG amplitude for the vibration condition was 22%, 30% and 48% higher for LONG-L1-L, LONG-L1-R and ILIO-L3-L, respectively, than for the no vibration condition. No *Time* effect was observed between the four different 15-min time blocks over the 60 min exposure for any of the muscles (Figure 5.3). Furthermore, no significant interactions were found. Depending on the muscle and condition, the average muscle activity was at or below 15.1% MVE for 90% of the total measurement period.

Table 5.3: Average APDF EMG amplitude values (% MVE) for both conditions and each 15-min cycle and *P* values from the corresponding 2-way ANOVAs

	%ile	Vibration				No Vibration				P value		
		1[a]	2	3	4	1	2	3	4	C[b]	T[b]	C × T[b]
LONG-L1-L	10	4.1	3.9	3.6	3.6	3.2	3.3	3.4	3.3	0.24	0.99	0.93
	50	6.2	5.6	5.3	5.3	5.3	4.1	4.2	4.3	0.06	0.58	0.98
	90	10.3	10.3	10.6	11.1	8.9	8.2	9.0	6.9	**0.04**	0.96	0.82
LONG-L1-R	10	6.2	5.6	4.9	4.7	4.0	4.0	4.1	4.0	**0.03**	0.86	0.82
	50	8.7	8.3	7.1	6.6	7.1	6.2	6.4	5.1	0.17	0.58	0.97
	90	13.9	15.1	14.5	13.5	10.5	9.5	9.9	9.9	**0.02**	1.0	0.97
ILIO-L3-L	10	4.8	4.8	4.7	4.2	3.2	2.9	3.3	3.2	**0.04**	0.99	0.97
	50	6.3	6.5	6.4	5.8	4.1	3.6	4.5	3.9	**0.02**	0.98	0.98
	90	13.3	10.0	10.5	9.5	5.7	5.3	6.3	5.9	**0.01**	0.87	0.87

%ile: percentile

[a] 15-min cycles (1: 0-15 min; 2: 16-30 min; 3: 31-45 min; 4: 46-60 min)

[b] C : Conditions (Vibration vs No Vibration); T: Time (between 15-min cycles): C × T: Condition × Time

Significant differences ($p \leq 0.05$) are identified with bold characters

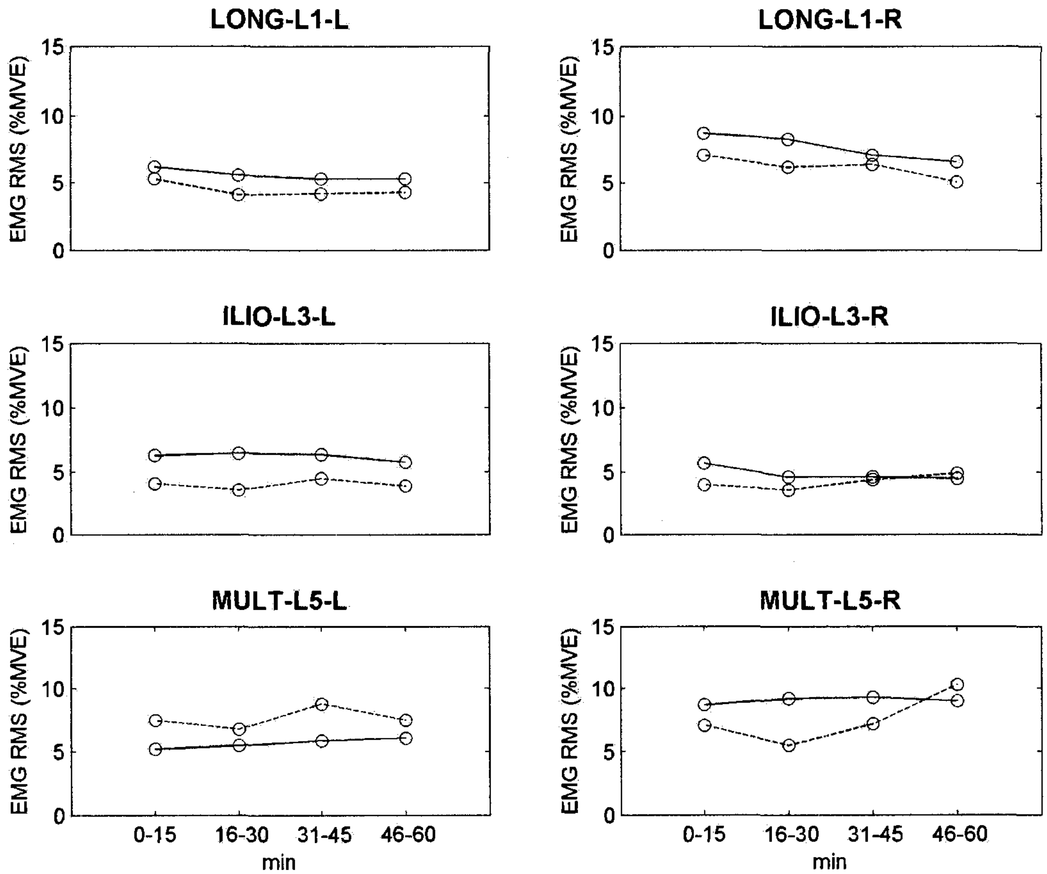


Figure 5.3: Median (50th percentile) APDF EMG amplitude values (% MVE) of the back muscles for each 15-min cycle during the 60-min exposure
Solid lines: Vibration; Dashed lines: No-vibration.

5.4.3 Balance

No statistically significant main *Condition* effects or *Condition* × *Period* interactions were observed for any of the COP summary measures (Figure 5.4). A *Period* effect was observed for all (except MFREQ_AP) summary measures reported. Post hoc comparisons revealed that only one summary measure (MFREQ_ML) differed significantly between PRE and POST, being smaller at the post assessment (Figure 5.4).

There was a tendency for the values during the recovery period (RECOV20, 40 and 60) to be smaller than PRE, and to get smaller as the recovery time increased. This tendency was also observed for MVELO, MPF_ML (Figure 4) and to a lesser extent with AREA_SW (not shown).

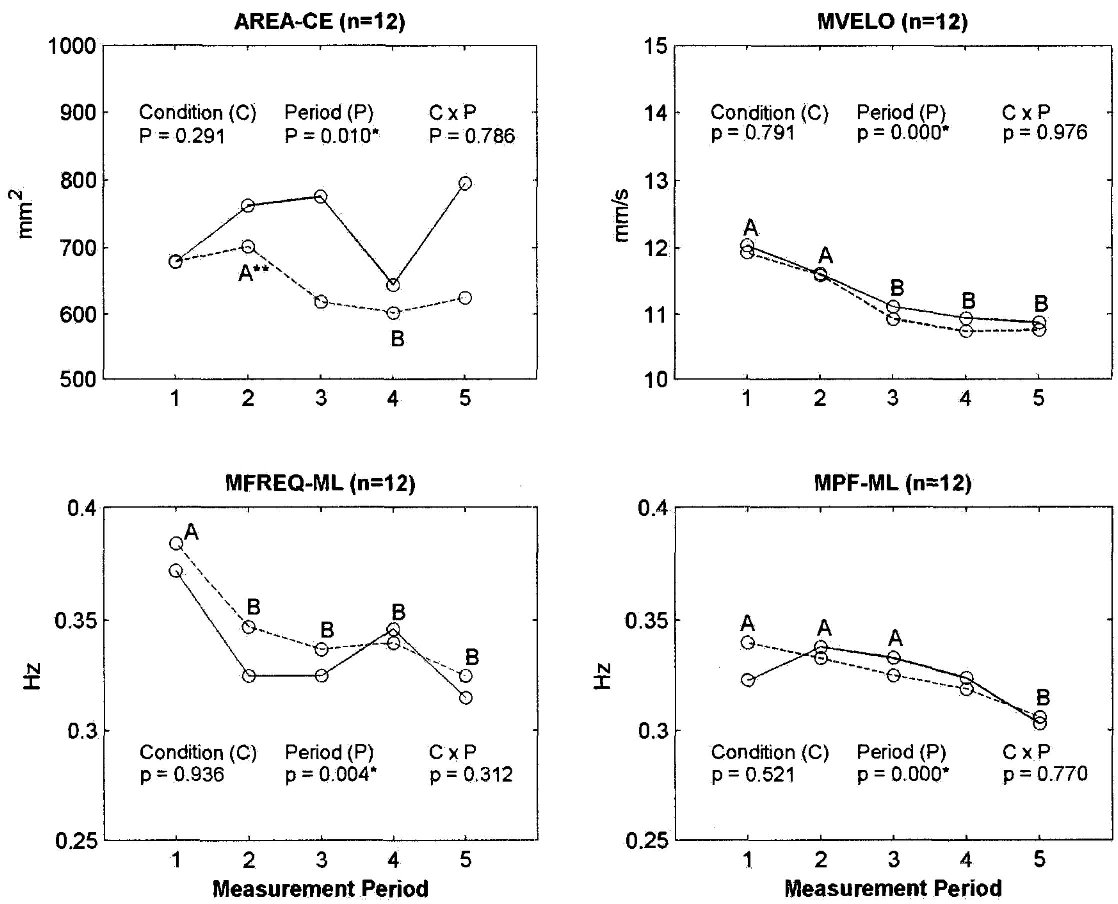


Figure 5.4: Average values for AREA_CE (mm^2), MVELO (mm/s), MFREQ_ML (Hz), and MPF_ML (Hz) for both sitting conditions at each measurement period (1: PRE, 2: POST, 3: RECOV20, 4: RECOV40, 5: RECOV40).
Solid lines: Vibration; Dashed lines: No-vibration.
ANOVA results (probability values) are displayed for the *Condition* and *Period* main effects, as well as their interaction.
* Significant differences (P≤0.05).
** The letters represent the results from Post hoc test (Bonferroni) on the *Period* effect.
The measurement periods represented by the same letters are not significantly different.

5.4.4 Reflex response

For the EMG reflex latency, no *Condition* × *Period* interaction effects were found for any of the muscles (Figure 5.5). Significant main effects of *Condition* and *Period* were found for only one muscle (LONG-L1). The *Condition* effect showed that the EMG latency for this muscle is higher during the vibration condition (78.3 ms) than during the no vibration condition (73.4 ms). *Post-hoc* comparisons showed that PRE measures were significantly higher than POST (79.8 versus 73.0 ms, respectively) and RECOV40 (74.5 ms) irrespective of the condition.

Regarding the EMG reflex amplitude ratio, the only significant main effect was with the *Period* for ILIO-L3 (Figure 5.5). *Post-hoc* multiple comparisons indicated that there were significant differences between PRE and POST and between PRE and RECOV60. During the vibration condition, average EMG reflex amplitude ratios ranged between 5.8 and 6.9, 6.7 and 8.3, and between 4.9 and 5.6 for LONG-L1, ILIO-L3, and MULT-L5, respectively. Similarly, during the no vibration condition, these values ranged between 5.7 and 7.4, 8.1 and 9.1, and between 5.7 and 7.5 for LONG-L1, ILIO-L3, and MULT-L5, respectively.

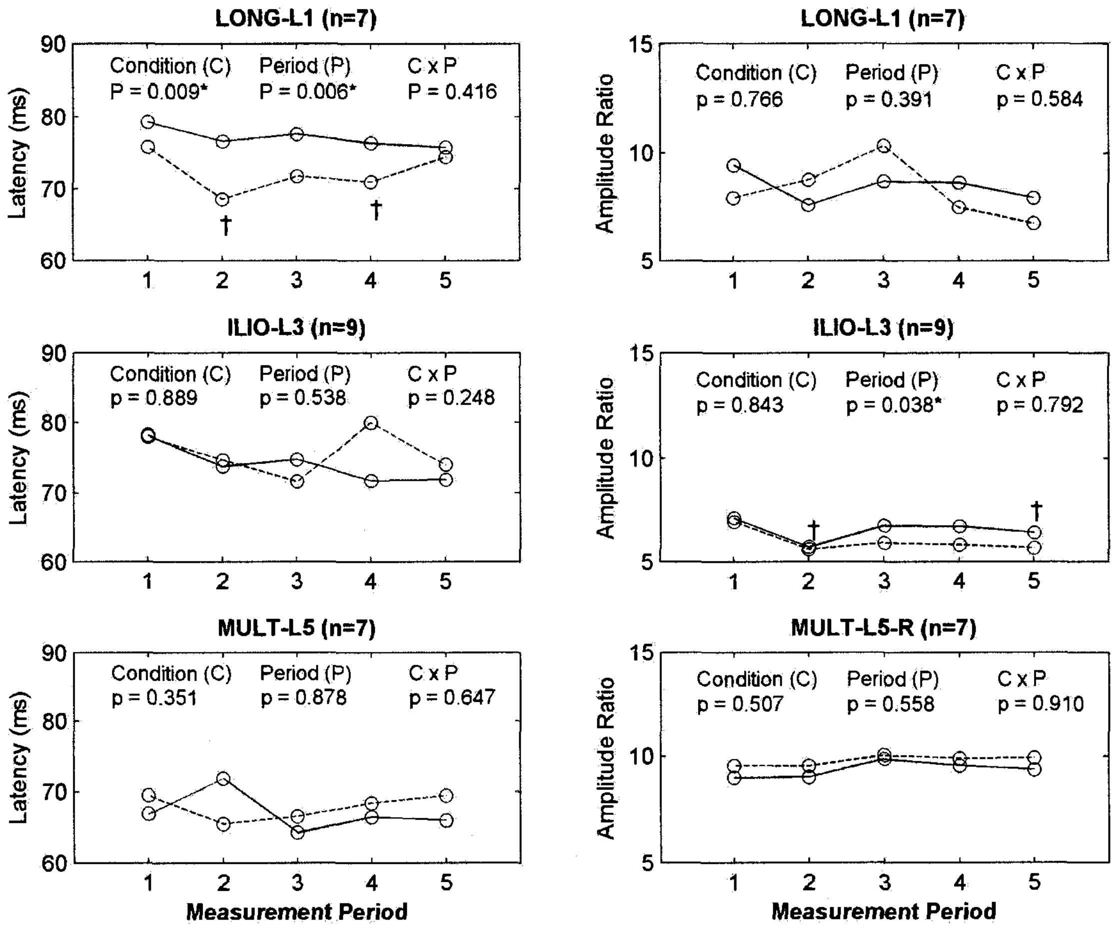


Figure 5.5: Average EMG reflex latency (ms) (left plots) and EMG reflex amplitude ratio (right plots) for both sitting conditions at each measurement period (1: PRE, 2: POST, 3: RECOV20, 4: RECOV40, 5: RECOV40).
Solid lines: Vibration; Dashed lines: No-vibration.
ANOVA results (probability values) are displayed for the *Condition* and *Period* main effects, as well as their interaction. Significant differences ($P \leq 0.05$) are indicated by an *.
†: significantly different from PRE

5.4.5 Muscular fatigue

In general, the average MPF decreased in all the muscles after the 60-min sitting exposure regardless of the vibration condition (no significant *Condition* × *Period* interactions were found), though a significant main effect of *Period* was obtained in three of the muscles (LONG-L1-L, ILIO-L3-L, ILIO-L3-R) (Figure 5.6). A significant effect for *Condition* was obtained only for MULT-L5-L where the average mean MPF was 133.8 Hz and 141.7 Hz for Vibration and No Vibration conditions, respectively.

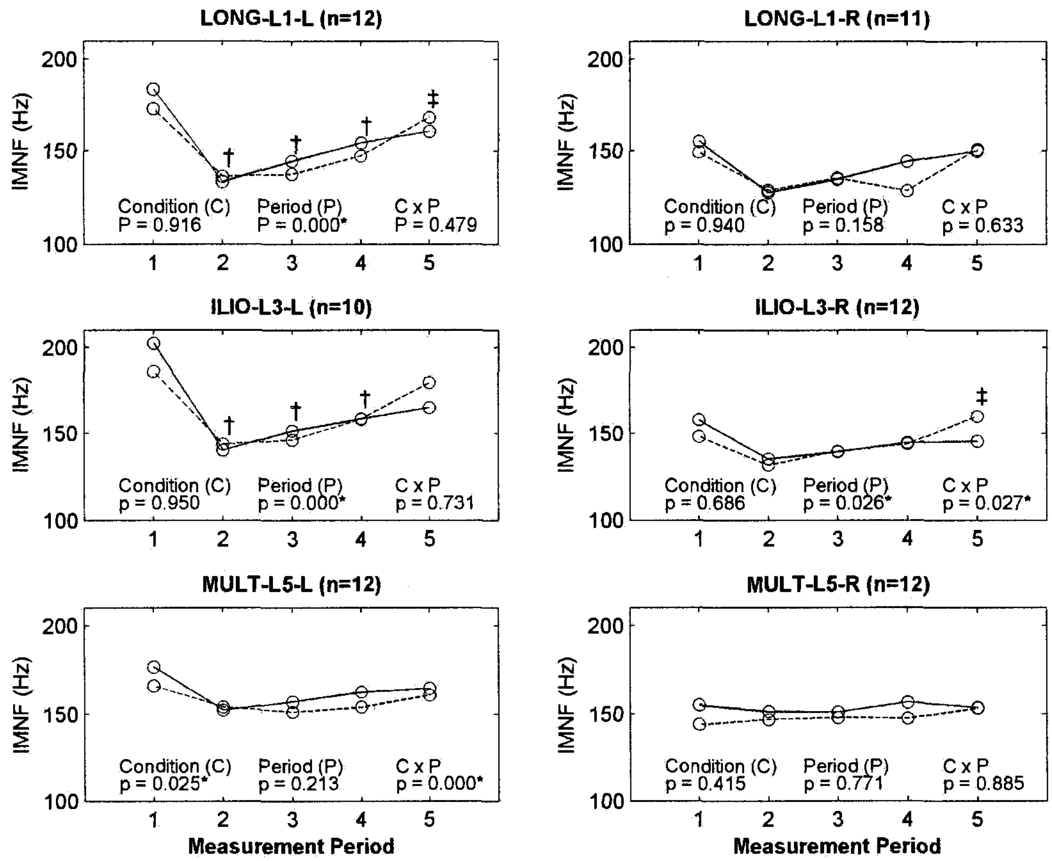


Figure 5.6: Average IMNF values for both sitting conditions at each measurement period (1: PRE, 2: POST, 3: RECOV20, 4: RECOV40, 5: RECOV40).
Solid lines: Vibration; Dashed lines: No-vibration.
ANOVA results (probability values) are displayed for the *Condition* and *Period* main effects, as well as their interaction. Significant differences (P≤0.05) are indicated by an *.
†: significantly different from PRE
‡: significantly different from POST

5.5 Discussion

The purpose of the present study was to test whether any acute effects on the sensorimotor system (balance, back muscle activity and fatigue, and back muscle reflex response) exist following seated vertical WBV exposure. Despite considerable measurement precautions taken to standardize the different tests and to maximize their reliability, WBV corresponding to mining vehicles combined with relevant body postures elicited very few effects on the measured variables. Overall, these results do not support our main hypotheses suggesting that exposure to WBV, when compared to a no WBV (i.e., sitting only) exposure, decreases balance, delays the EMG reflex latency, decreases the EMG reflex amplitude, and induces higher muscle activity and fatigue. However, new findings have emerged concerning the effect of sitting per se regarding muscle fatigue and balance.

Before discussing on the effect of WBV on the different variables, it is important to demonstrate that the conditions of exposure were comparable between the four testing sessions. Effectively, it is easy to control the vibration exposure using the WBV simulator but it is more difficult to control the back posture. Mean lumbar angles were found to be comparable as well as the mean back muscle activity (% RVE) within the same condition of exposure giving support to the appropriateness of our standardization procedures.

5.5.1 Muscular activity

On average, exposure to WBV significantly increased the level of muscle activity for LONG-L1-L, LONG-L1-R and ILIO-L3-L by 22%, 30% and 48%, respectively. To the authors' knowledge, this is the first study to show an increase of back muscle activation due to WBV, comparatively to a control sitting-only condition. If we look at prolonged sitting (without vibration), our values were slightly higher than previous findings (Callaghan and McGill, 2001; Durkin et al., 2006), most likely due to the fact the subject's trunk was always bent forward without resting on the backrest. Overall, the average EMG (50th percentile APDF) was below 15% MVE during the vibration condition, as observed in a group of helicopter pilots (de Oliveira and Nadal, 2004).

One possible explanation for the increase in muscle activity could be the involvement of the "tonic vibration reflex" (TVR). This reflex is caused by vibratory activation of the primary endings of the muscle spindles, which are muscle receptors sensitive to stretch. However, this may not have been the case in the current study as the TVR is induced by vibration at higher frequencies than induced in the present study (0.5-20 Hz), namely in the 30-120 Hz frequency band (Vermeersch.et al., 1986). Alternatively, the increased muscle activity could be attributed to the need to counteract the bending moment generated by WBV and to co-contraction of the abdominal and back muscles, the latter being the response to the need for increased spinal stability (Granata and Orishimo, 2001).

As stated in the introduction, it is of interest to see whether the increase of muscular activity observed during vibration could lead to muscular fatigue. However, the muscular activity was quite low for all muscles after sitting for 60 min, whether subjects were exposed or not to WBV. If we use the thresholds of 5% MVE, 14% MVE and 50% MVE (Jonsson, 1978) for the 10^{th}, 50^{th} and 90^{th} percentile APDF, respectively, only in a few cases were these thresholds exceeded (proposed to avoid muscular fatigue over an eight-hour period), though not by very much. Also, the non-significant *Period* and *Condition* × *Period* interaction (EMG amplitude APDF analysis) do not support this hypothesis.

On the other hand, the amplitude of EMG is not a reliable index of muscle fatigue (Lariviere et al., 2002; Nargol et al., 1999) especially when the postures and muscle efforts are not strictly controlled such as in the present study. The PRE and POST exposure back muscle contractions performed during the back muscle reflex test, combined with spectral analyses that leads to more reliable results (Lariviere et al., 2002; Nargol et al., 1999), were more suited to address this issue, as discussed in the next section.

5.5.2 Muscular fatigue

Muscular fatigue is often manifested by a shift of the power spectrum toward lower frequencies (Dimitrova and Dimitrov, 2003). This study demonstrated significant decreases in MPF for three muscles (LONG-L1-L, ILIO-L3-L, and ILIO-L3-R) after 60 minutes of sitting for both vibration and no vibration conditions. Interestingly, two of these muscles were also shown to be the only ones to have significantly increased muscular activity due to WBV, as discussed earlier. However, the *Condition* × *Period* interaction was not significant in any of these muscles. Thus, we cannot speculate that WBV alone elicited muscle fatigue. It would seem that the sitting task per se generated this fatigue.

To the author's knowledge, only one (Hansson et al., 1991) of three studies (de Oliveira and Nadal, 2004; El Falou et al., 2003; Hansson.et al., 1991) have observed back muscle fatigue during sitting tasks combined with WBV, as assessed with spectral analyses of EMG signals. It should be noted that Hansson's subjects maintained a constant trunk flexion of 20º and wore a 4 kg weight to load their trunk, these to induce fatigue and not necessarily a realistic situation of workers exposed to WBV. On the other hand, these results show that fatigue occurs when an increased level of muscular activity is necessitated, as probably occurred in the present study. It could possibly be that muscle fatigue during sitting is highly task dependent. For example, in this study, subjects were required to maintain 15° of trunk flexion (i.e., they were not permitted to use the backrest). This was in addition to maintaining varying degrees of neck rotation. Thus, the present findings showed that back muscle fatigue during sitting is induced in situations where an active posture is maintained. Obviously, these results might only be generalized to driving conditions necessitating these postures, such as in mining vehicles where visual inspection of the environment, in front or on the sides of the vehicles, is required to avoid accidents.

5.5.3 Balance

Again, the present results demonstrated that WBV did not have any effect on balance. This concurs with Cornelius et al. (1994), though not with others (McKay, 1972; Seidel et al., 1980). Cornelius et al. (1994) has suggested that the effect on balance may depend on three parameters: (1) the frequency, (2) the duration and (3) the direction of the vibration exposure. Regarding the frequency of vibrations, the current study and that of Cornelius et al.(1994) studied vibration exposure representative of real vehicles (more random) while the other studies (McKay, 1972; Seidel.et al., 1980) used sinusoidal vibrations. Regarding the duration of exposure, it was considerably longer (180 min) in Seidel et al. (1980) than in the other studies (≤ 60 min). Finally, regarding the direction of vibrations, all the studies used vertical vibrations. Interestingly, vibration in the vertical direction has previously shown to affect the vestibular system (Suvorov et al., 1989), one of the three major sensory systems involved in balance.

There was a significant *Period* effect indicating a decrease of various COP summary measures following the 60-min exposure. Unexpectedly, these results, although small in magnitude, mean an improvement of balance. Since there was no *Condition* × *Period* interaction, the effect of vibration was rejected, thus leaving sitting exposure the cause of this improvement. An interesting observation was that effects on the COP measures continued during the recovery period. Between each series of measures during the recovery period, subjects were instructed to sit in a chair but without using the backrest. Therefore, once again, the subjects were exposed to a sitting posture.

Even more unexpected is an improved balance with back muscle fatigue, which is in contradiction with previous findings (Davidson et al., 2004; Madigan et al., 2006). One reason we may have seen different results is that the level of back muscle fatigue reached in the current study was probably much lower. Back muscle fatigue was generated in these studies (Davidson et al., 2004; Madigan et al., 2006) so that the MVCs reached as low as 60% of the pre-fatigued MVC. It is doubtful that these levels of fatigue were reached in the current study considering the low activation level of back muscles recorded during the 60-min exposure.

A decrease of only 1 mm/s was observed for the MVELO variable (Figure 5.4), which is considerably smaller than the 6 mm/s increase observed earlier following intense back muscle fatigue (Davidson et al., 2004). Although small and possibly of questionable physiological significance, it is unlikely that the present findings results from a type-I error because many other COP summary measures (n = 26/36, from Prieto et al., 1996) not presented here showed a significant *Period* effect. These small differences were probably detected because several trials (n = 8) were averaged at each period thus increasing statistical power. At this moment, we are unable to identify what would explain these findings. There might be a learning effect due to repeating standing trials that may appear between the different sets of eight trials or there might be an unknown phenomenon acting during the sitting exposure. The reader is reminded that the possible learning (or fatigue) effect within the PRE and within the POST trials was rejected, thus justifying the averaging of the eight standing trials.

Poor postural balance is cited as being associated with a higher risk of falling (Piirtola and Era, 2006), especially in the elderly. However, since falls in workers are more likely to occur while egressing from a vehicle or from slipping due to uneven or wet terrain, our use of static posturography (or postural steadiness) to measure the risk of falling from dynamic situations may be questioned. Because of the simplicity of this type of measurement, the characteristics of the movement of the COP from a single platform are a common and reliable outcome measure in a vast majority of research in quiet standing (Winter, 1995). Future studies could possibly look at the effects of balance while egressing a vehicle or possibly looking at balance during perturbed standing or during gait.

Laboratory experiments do not usually afford researchers to fully simulate environment factors. One such factor is noise, a condition found in the mining environment, which has been shown as a detrimental factor effecting postural sway (Juntunen et al., 1987). In a series of studies with different combinations of noise and WBV and other environmental factors, though the effects varied in a non-systematic, the combination of both noise and WBV had a strong tendency to increase body sway amplitudes.

5.5.4 Reflex response

Exposure to WBV, when compared to sitting alone, did not have any significant effects on the EMG reflex latency or the EMG reflex amplitude values (no significant *Condition* × *Period* interaction), contrary to previous findings (Wilder et al., 1996). Interestingly, in the sitting condition (without vibration), the response significantly decreased from 107 ms to 89 ms. This was the same trend, though not statistically significant, in the current study. Some significant *Condition* and *Period* main effects were observed here but these effects were rather inconsistent across muscles and periods of measurements. It seems thus preferable to not speculate further on these findings.

The values of our reflex latencies were lower than what were measured by Wilder et al. (1996). The most likely explanation for these differences is the way in which the sudden load was applied. The subjects in Wilder et al. (1996) stood upright and held an instrumented pan in their hands to catch a falling tennis ball. The subjects in our study had their pelvis fixed and had the load directly applied to their trunk, bypassing the arms altogether. Finally, Wilder et al. (1996) found that walking for five minutes after vibration exposure reduce the reflex response. This effect could have influenced our results. Our subjects had to descend the vibration simulator and then walk a short distance to and be fixed into the sudden loading apparatus. This delay after the end of the 60 minutes of sitting and the sudden loading trials could have masked any potential effects, though it would have been systematic for both vibration conditions. If this is the case, we can question the physiological relevance of such an effect on the risk of injury after exposure to WBV.

5.5.5 Study Limitations

The present study suffers from several limitations. The results could potentially differ if the various parameters of WBV exposure are different: duration and direction (horizontal), seat characteristics, frequency and amplitude characteristics. Therefore, there is a potential for future studies to investigate these parameters either alone or in a combination. Also, the subjects' characteristics were relatively homogeneous relative to age, sex and body mass index, which may affect the response to WBV.

5.6 Conclusion

The current research provided a thorough investigation of the possible acute effects of WBV on various aspects of the sensorimotor system that may affect the stability of the spine. In previous studies investigating the effects of WBV, a control (or no vibration) condition has not been utilized for comparison, and thus difficult to conclude. The results of this study demonstrated that exposure to WBV elicits significantly higher, though low-level, back muscle activity than sitting without vibration. Muscle fatigue of the longissimus and iliocostalis lumborum muscles as well as some variables associated with balance were significantly affected after sitting for 60 minutes, however, WBV alone did not induce effects any more than sitting without vibration. This emphasizes that WBV per se is not necessarily responsible for such acute effects. Sitting without vibration appears to have the potential to influence back muscle fatigue and postural balance. However, this may only be attributed to the constrained trunk posture adopted during the 60-min of exposure and to the vibration exposure typical of LHD operators.

Acknowledgements

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CHAPTER 6

SENSITIVITY OF DIFFERENT TRUNK BIOMECHANICAL RESPONSES TO SEATED WHOLE-BODY VIBRATION

This chapter describes the methodology and reports the results related to the second study (Study 2) of this dissertation.

6.1 Methods

The methodology of this second study follows very similarly to the methodology outlined in Chapter 5. This section will only highlight and describe any differences between the two studies.

6.1.1 Subjects

Twelve healthy males were recruited to participate. Participant mean ± SD age, height and weight were 25 ± 7 years, 1.8 ± 0.1 m, and 78 ± 9 kg, respectively. Exclusion criteria is the same as described in section 5.3.1. Before their participation, all subjects were informed of the experimental protocol and of its potential risks, and signed a consent approved by the university Research Ethics Committee.

6.1.2 Study design, tasks and general procedures

6.1.2.1 Study design

A repeated-measures design was used to study the effect of a 60-min exposure (vibration/no-vibration) on six biomechanical measures using three different tests (balance/sudden loading/stadiometry). Participants performed a total of six experimental conditions (3 tests [balance/sudden loading/stadiometry] × 2 exposures [vibration/no vibration]), each on a separate day, with at least one day and no more than one week between testing sessions. The conditions were presented to the subjects according to a counterbalanced design. For each test, measurements were taken before (PRE) and immediately after (POST) the 60 minutes of exposure. During the (with) vibration condition only, additional measurements were taken at 20 (RECOV20), 40 (RECOV40), and 60 (RECOV60) minutes after the end of exposure to evaluate if there were any possible effects that remained after the exposure (recovery).

6.1.2.2 Procedures

Please refer to the illustration of the measurement protocol presented in Figure 5.1. This study follows the same protocol (with the only difference being that measures to evaluate recovery were only taken during the vibration condition, as stated in the preceding section).

<u>Reference contractions, postural balance test, and sudden loading test</u>

Please see section 5.3.2.2

Stadiometry

A stadiometer (Figure 6.1) was used to measure spinal height variation. The stadiometer is a metal structure slightly tilted backward (15°) that offers the subject feedback to control his upright posture. Four pressure transducers are mounted on the base to measure the weight distribution over the heels and soles and over the left and right foot. An inclinometer mounted on a pair of plastic eyewear was used to control head tilting. The force pressure transducers are positioned to force the subjects to extend the knees against padding, which is height adjustable. The backrest also carries supports that are adjustable in height. The lower part of the backrest is a support for the buttocks and an adjustable rod is also attached that is positioned in the curve of the lower back. This rod protrudes forward and backward and is used to control the posture of the lower back. Furthermore, there is another support at the upper thoracic spine and another support for the head. There is another adjustable rod that is positioned behind the curve of the neck, which is also used to control the posture. Visual feedback on a computer monitor directly at eye-level in front of the subject was used to help him in reproducing their weight distribution and posture in subsequent measurements.

For each trial, the subject stepped up onto the stadiometer and adjusted his posture with his arms crossed in front of his chest. Once the correct posture had been achieved, the experimenter instructed him to take a deep, but not forceful, inhalation and then exhalation. The measurement was taken at the end of the exhalation. The measurement of stature was made by aligning a horizontal laser beam with a landmark at the level of the seventh cervical vertebra (C7). A high precision potentiometer was used to measure the position of this laser beam. After each trial, the subject stepped off the stadiometer and stepped back on to repeat the same procedure. A total of five trials were made.

Seated exposure

During the 60 min of seated exposure (with or without vibration), participants were seated on a hard, rigid seat with no backrest (Figure 6.2), that was mounted on the vehicular vibration simulator. The vibration exposure to which the subjects were exposed ranged between 0.5 and 20 Hz, with the dominant frequency at 4.5 Hz, and a frequency-weighted average acceleration ($a_{w(0.5\text{-}20Hz)}$) of 1.4 m/s^2. During the 60-min exposure, the subject placed his hands on his lap, were asked to maintain their upper body an erect and straight (but relaxed), and to look forward at all times.

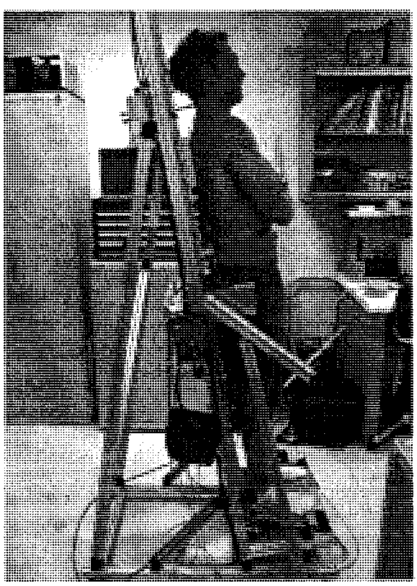

Figure 6.1: Stadiometer

Figure 6.2: Rigid seat on whole-body vibration vehicular simulator

6.1.3 Measurement techniques

All signals were collected, converted to digital signal via a 16-bit A/D converter, and stored on a hard disk for later analysis. Please refer to sections 5.3.3.1, 5.3.3.2, and 5.3.3.3 for details regarding EMG, the force plate and goniometer, respectively.

6.1.4 Data processing

All data processing and data reduction were performed using MATLAB (Version 7.0, The Mathworks, Natick, MA). The methods by which the different outcome measures were computed are described in detail in section 5.3.4.

6.1.5 Statistical analysis

Data were analyzed using the NCSS software (Number Cruncher Statistical Systems, Kaysville, Utah, USA). The same statistical analyses were performed as described in section 5.3.5. The only difference being that rather than 5 measurement periods being assessed (i.e., PRE, POST, RECOV20, RECOV40, RECOV60), only 2 measurement periods (PRE and POST) were used in the ANOVA to assess the differences between the two vibration conditions and the measurement periods and its interaction for the COP summary measures, reflex latency and amplitude and IMNF.

6.2 Results

6.2.1 Muscular activity

Significant main effects for *Condition* were found for two muscles (Table 6.1: ILIO-L3-R and MULT-L5-R). The EMG amplitude for the vibration condition was 33% and 71% higher for ILIO-L3-R and MULT-L5-R, respectively, than the no vibration condition. A *Time* effect was observed between the four different 15-min time blocks over the 60-min exposure for MULT-L5-L for the 90^{th} %ile $APDF_{MVE}$ (Table 6.1). A significant *Condition* × *Time* interaction was found for MULT-L5-R for the 10^{th} %ile $APDF_{MVE}$ (Table 6.1). Depending on the muscle and condition, the average muscle activity was at or below 25.4% MVE for 90% of the total measurement period.

6.2.2 Balance

No statistically significant *Condition* effects were observed for any of the COP summary measures (Table 6.2). A *Period* effect was observed for both AREA_CE and MFREQ_ML, with the POST values decreasing after 60 minutes of exposure. No *Condition* × *Period* effects were observed.

6.2.3 Reflex response

For the EMG reflex latency, no *Condition* effects or *Condition* × *Period* interaction effects were observed for any of the muscles (Table 6.3). A significant *Period* main effect was shown for ILIO-L3-R. The EMG latency for this muscle showed the PRE measures were significantly longer (higher) than the POST measures.

For the EMG reflex amplitude ratio, there were neither any *Condition* nor *Period* main effects, nor any *Condition* × *Period* interaction effects for any of the muscles (Table 6.4).

6.2.4 Muscular fatigue

No significant *Condition* main effects, or any *Condition* × *Period* interaction effects were observed for any of the muscles (Table 6.5). Significant main effects of *Period* were observed in two muscles (LONG-L1-L and MULT-L5-L). In all cases, the average mean power frequency (MPF) values increased after the 60-min sitting exposure regardless of the vibration condition (Table 6.5).

6.2.5 Stadiometry

No differences in spinal shrinkage were observed after 60 minutes of exposure to either of the vibration conditions (Table 6.6).

Table 6.1: Average APDF EMG amplitude values (% MVE) for both conditions and each 15-min cycle and *P* values from the corresponding 2-way ANOVAs

	%ile	Vibration				No Vibration				P value		
		1[a]	2	3	4	1	2	3	4	C[b]	T[b]	C×T[b]
LONG-L1-L	10	9.6	9.3	8.9	9.5	10.2	8.1	7.0	6.9	0.256	0.362	0.824
	50	15.1	15.0	14.5	16.3	14.4	12.7	13.2	13.8	0.124	0.957	0.960
	90	23.0	22.9	22.8	25.1	21.0	21.4	21.8	22.3	0.126	0.957	0.972
LONG-L1-R	10	10.2	10.4	10.1	9.9	11.4	10.1	9.3	10.6	0.549	0.314	0.549
	50	16.8	16.5	16.4	16.5	15.7	15.7	15.0	16.4	0.404	0.658	0.772
	90	25.4	25.0	25.1	25.2	21.6	23.4	22.8	23.9	0.330	0.988	0.830
ILIO-L3-L	10	4.0	3.8	3.5	3.3	3.1	3.1	2.7	2.8	0.213	0.155	0.415
	50	5.1	4.8	4.8	4.5	3.9	3.8	3.9	3.8	0.193	0.861	0.426
	90	7.4	6.5	6.7	6.8	5.0	5.4	5.6	6.0	0.194	0.907	0.362
ILIO-L3-R	10	5.0	5.3	5.5	5.4	3.9	4.0	3.4	4.5	0.304	0.614	0.531
	50	7.4	7.9	8.5	8.7	5.5	6.6	5.2	7.0	0.287	0.364	0.593
	90	11.1	11.7	12.6	13.4	8.0	9.4	8.8	10.4	**0.048**	0.100	0.631
MULT-L5-L	10	3.7	3.6	3.4	3.9	3.3	3.5	3.2	3.2	0.399	0..404	0.540
	50	4.8	5.3	5.1	5.5	4.9	4.6	4.9	6.0	0.878	0.817	0.655
	90	7.5	7.5	8.5	10.3	6.6	8.5	8.4	9.9	0.918	**0.01**	0.376
MULT-L5-R	10	7.1	8.0	9.1	9.7	4.2	4.4	3.8	3.9	**0.002**	0.226	**0.025**
	50	8.7	11.5	11.3	12.2	6.6	6.3	6.2	6.6	**0.010**	0.131	0.077
	90	12.1	14.9	14.9	16.1	8.1	9.5	9.5	10.5	**0.021**	0.016	0.951

%ile: percentile

[a] 15-min cycles (1: 0-15 min; 2: 16-30 min; 3: 31-45 min; 4: 46-60 min)

[b] C : Conditions (Vibration vs No Vibration); T: Time (between 15-min cycles): C × T: Condition × Time

Significant differences (p≤0.05) are identified with bold characters

Table 6.2: Average values (and Standard Deviation) for AREA_CE (mm^2), MVELO (mm/s), MFREQ_ML (Hz), MPF_ML (Hz) for both sitting conditions at each measurement period and *P* values from the corresponding 2-way ANOVAs

	No Vibration		Vibration		P value		
	Pre	Post	Pre	Post	C [a]	P [a]	C × T [a]
AREA_CE	462 (307)	658 (458)	666 (532)	525 (411)	0.42	**0.00**	0.54
MVELO	10.3 (2.4)	10.4 (2.3)	10.7 (2.4)	10.1 (2.1)	0.78	0.31	0.14
MFREQ_ML	0.39 (0.11)	0.33 (0.11)	0.40 (0.12)	0.34 (0.12)	0.34	**0.01**	0.99
MPF_ML	0.33 (0.08)	0.34 (0.07)	0.34 (0.07)	0.33 (0.06)	0.88	0.54	0.07

[a] C : Conditions (Vibration vs No Vibration); P: Period (PRE vs POST): C × P: Condition × Period
Significant differences ($p \leq 0.05$) are identified with bold characters

Table 6.3: Average EMG reflex latency values (ms) for both sitting conditions, each measurement period and *P* values from the corresponding 2-way ANOVAs

	No Vibration		Vibration		p-value		
Muscle	Pre	Post	Pre	Post	C [a]	P [a]	C × T [a]
LONG-L1-L	82	78	89	81	0.30	0.10	0.41
LONG-L1-R	87	84	81	80	0.96	0.06	0.36
ILIO-L3-L	79	73	78	82	0.48	0.42	0.17
ILIO-L3-R	79	77	84	77	0.20	**0.05**	0.55
MULT-L5-L	75	72	66	69	0.28	0.59	0.16
MULT-L5-R	76	73	74	67	0.50	0.15	0.41

[a] C : Conditions (Vibration vs No Vibration); P: Period (PRE vs POST): C × P: Condition × Period
Significant differences ($p \leq 0.05$) are identified with bold characters

Table 6.4: Average EMG reflex amplitude ratio values for both sitting conditions, each measurement period and *P* values from the corresponding 2-way ANOVAs

	No Vibration		Vibration		p-value		
Muscle	Pre	Post	Pre	Post	C [a]	P [a]	C × T [a]
LONG-L1-L	8	8	7	7	0.28	0.45	0.58
LONG-L1-R	11	11	10	9	0.22	0.79	0.59
ILIO-L3-L	7	7	7	8	0.93	0.62	0.20
ILIO-L3-R	10	10	8	9	0.42	0.85	0.23
MULT-L5-L	10	10	8	7	0.07	0.64	0.61
MULT-L5-R	10	10	8	9	0.26	0.26	0.24

[a] C : Conditions (Vibration vs No Vibration); P: Period (PRE vs POST): C × P: Condition × Period
Significant differences ($p \leq 0.05$) are identified with bold characters

Table 6.5: Average IMNF values (Hz) for both sitting conditions, each measurement period and *P* values from the corresponding 2-way ANOVAs

	No Vibration		Vibration		p-value		
Muscle	Pre	Post	Pre	Post	C [a]	P [a]	C × T [a]
LONG-L1-L	121	157	132	163	0.49	**0.01**	0.81
LONG-L1-R	129	135	119	145	0.73	0.10	0.79
ILIO-L3-L	138	148	130	156	0.48	0.07	0.59
ILIO-L3-R	126	140	115	139	0.62	0.10	0.65
MULT-L5-L	135	149	138	165	0.23	**0.01**	0.43
MULT-L5-R	136	146	132	140	0.46	0.24	0.87

[a] C : Conditions (Vibration vs No Vibration); P: Period (PRE vs POST): C × P: Condition × Period
Significant differences ($p \leq 0.05$) are identified with bold characters

Table 6.6: Spinal shrinkage after 60 minutes of exposure

	Average (mm)	SD (mm)	Mininum (mm)	Maximum (mm)
Without Vibration	3.5	1.0	2.2	5.7
Vibration	3.6	1.4	1.5	6.1

CHAPTER 7

GENERAL DISCUSSION

The general goal of this study was to evaluate the trunk biomechanical responses to seated WBV. The mechanism(s) by which WBV contributes to the etiology of low back disorders and injuries are not well understood. We proposed that one potential mechanism is that WBV effects spinal stability. Using Panjabi's (1992) model of spinal stability as the conceptual framework, several biomechanical measures were used to investigate the effects of WBV on the active, passive and neuromuscular control subsystems of this model. Therefore, to reach the overall objective of the study, the work arising from this dissertation focused on investigating the: 1) reliability of the different biomechanical measures that could potentially be used to assess the different subsystems in Panjabi's spinal stability model; and 2) effects of WBV on the three subsystems. These were investigated in three separate studies; the first being to investigate the reliability of the biomechanical measures (Chapters 3 and 4) and the second and third being to investigate acute effects of WBV (Chapters 5 and 6). Considering that more specific issues have been already discussed in each chapter, the present general discussion will focus on broader issues related to the reliability study and to the comparability of findings from studies 2 and 3. Additionally, the present findings need to be compared with more recent findings that were published after the present studies were initiated.

7.1 Reliability of the different biomechanical measures

Study 1 addressed the first research question, "Are the proposed biomechanical measures reliable?" Although data of biomechanical measurements such as the COP, reflex response, and stadiometry are frequently reported, there is definitely a lack of standardization of methodology. For example, in the case of COP, the summary measures that are reported vary from one study to another. In another example, the way in which the reflex response is determined (i.e., the calculation of the onset time) also varies or may not even be reported. The purpose of this study was to perform a systematic investigation of the reliability of various biomechanical measures using the Generalizability Theory (Shavelson & Webb, 1991).

The Generalizability Theory (G-Theory) goes beyond the intra-class correlation coefficient (ICC) by examining the error variances using analysis of variance (Brennan, 2001; Shavelson & Webb, 1991). The ICC can be considered a special case of G-Theory. The G-theory can be used to determine measurement strategies (i.e., the number of trials needed) or assist in designing the measurement procedure. In G-Theory, reliability is based on the defined universe, and the specific components on error terms is quantified. This is a more flexible approach to the assessment of reliability than traditional methods. It is suggested that reliability should not be viewed as a property of the test but rather as a set of scores associated with testing procedures (Brennan, 2001) and that reliability can be different for different objects of measurement and different universes of generalization. Thus, reliability or standard error of measure is dependent on the sources of error that are incorporated into the measurement design. This implies that there is no definitive way to identify what sources of error should be included in the analysis. To achieve a reliable measure, it is very likely that multiple measurements would be needed to obtain a stable score. Therefore, averaging of multiple scores will always be more stable than a single measurement.

7.1.1 Summary of the reliability of balance, reflex response and stadiometer measures

This study found that the majority of the measures displayed poor to moderate reliability. For the COP summary measures, the majority of the summary measures achieve acceptable when at least seven trials are averaged during the same testing day. The same trend was true for the reflex response measures. With this, at least the minimum acceptable reliability will be reached.

For the stadiometer, the ICC was 0.3 and SEM was 1.5 mm. It was very difficult to achieve repeatable results over five measures. When five measures were taken, the standard deviation of these measures ranged from 0.26 to 3.5 mm. Some subjects obviously had difficulty reproducing their posture from one trial to another.

A more recent study has examined postural stability specific to the trunk (Lee & Granata, 2008). Depending on the outcome measure, they found the ICC values to be poor to moderate for the both intra-session (range: 0.26-0.93, mean: 0.60) and inter-session (range: 0.13-0.84, mean: 0.47) reliabilities. Therefore, this could be an alternative method to use rather than standing balance, which was used in this current research. Likewise, Moorhouse and Granata (2007) used an alternative method to quantify the reflex response of the torso musculature. Anterior-directed trunk perturbations of ± 2mm could be applied at a rate of 50 pulses per 30-s trial. Although the reliability of this method has yet to be determined, this type of method would obviously ease the increase of the number of trials. Therefore, it could be worthwhile to look at these alternative measurement techniques for future studies.

7.1.2 Limitations

The sample size may be the most noticeable limitation. Some researchers suggest anywhere from 8 participants (Hopkins, 2000) to 400 participants (Charter, 1999). This study had a sample size of 15 participants. Furthermore, depending on the measurement technique, certain subjects were omitted from the analyses due to technical problems (i.e., only 12 subjects were kept for the COP reliability analysis). These numbers may not have provided enough power for the reliability study. Depending on the measurement technique, previous reliability studies have reported the participation of between 7 and 49 subjects.

Another limitation is that the subjects who participated in this study were young and healthy adults. This is a homogenous sample and thus, the inter-subject variability might be low, and thus, leads to low ICC values. However, the ICC is not the only index of reliabililty. The SEM could also be an index of interest especially if the measure has the potential to discriminate between subjects (between-subject designs) and the SEM for test-retest studies (within-subject designs).

7.2 The effects of whole-body vibration

The second study addressed the research question, "Which biomechanical measures and variables are most likely to be sensitive in detecting responses due to WBV exposure?" and the third study addressed the research question, "Does exposure to conditions typically found in the workplace have an effect on the biomechanical responses to seated WBV?" The results from these two studies indicate that WBV per se is not responsible for any effects that were found. This lack of effect of WBV could, however, be due to the lack of sensitivity of the measures used. Even though two different vibration exposure profiles, postures and seats were used in both studies, the results obtained were similar. Thus, knowing this and the fact that WBV per se was not responsible for the effects, it is possible that sitting itself for 60 minutes could elicit effects. The results of the third study are thoroughly discussed in Chapter 5 (Santos et al., 2008).

7.2.1 Muscular activity

For muscular activity, certain muscles exhibited higher muscular activity during the vibration condition than the no-vibration (sitting only) condition. The average EMG (50^{th} %ile APDF) values appear to be in the same range for both studies, with the exception of the longissimus muscles (LONG-L1-L and LONG-L1-R) having slightly higher average EMG values in the second study. There was one interaction effect found in the second study (10^{th} %ile APDF for MULT-L5-R). One could say that the 10^{th} %ile APDF for this muscle is sensitive to WBV under the conditions at which we exposed our subjects. However, this finding should be interpreted with caution as it was the only significant result found. This could simply be a Type I error (spurious finding).

7.2.2 Muscular fatigue

For muscular fatigue, both studies showed no vibration effect, although there differences in the trends found over time. In the third study, the MPF values systematically decreased after the 60-min exposure for all muscles in both vibration and no-vibration conditions thus indicating the presence of back muscle fatigue. However, in the second study, the MPF values increased after the exposure. This discrepancy in results could be possibly due to two reasons. First, the postures that the subjects adopted in both studies were very different. In study 2, the subjects adopted a symmetric, forward-facing trunk postures. In study 3, the subjects adopted asymmetric, rotated trunk postures. It must be remembered, however, that even though asymmetric postures were adopted in the third study, main differences between studies 2 and 3 were the different seats used and vibration exposure simulated. Therefore, these trunk rotated postures could be one possible responsible for the back muscle fatigue observed in the latter study. The second reason being our method of analyzing muscular fatigue. In both studies, depending on the muscle and condition, the average muscle activity was at or below 25% MVE for 90% of the 60-min measurement period. It is debated if, during low-force contractions, muscles actually do fatigue since only minor and subtle physiological changes do occur. Although EMG is often used to measure fatigue, the results in studies with low-force contractions are inconclusive (Arendt-Nielsen, Mills, & Forster, 1989; Hansson et al., 1992). There are limitations of using EMG spectral analysis as a method of detecting localized muscle fatigue during low-level contractions. The mean power spectral frequency does not always reveal a decreasing trend in the motor unit conduction velocity. The recruitment of a new motor unit during contraction may result in an increase or a decrease of characteristic spectral frequencies (Farina, Zennaro, Pozzo, Merletti & Läubli, 2006).

Another promising method for detecting muscle fatigue in low-force contractions is with the mechanomyogram (MMG) (Barry, Geiringer, & Ball, 1985; Dalton & Stokes, 1993; Herzog, Zhang, Vaz, Guimaraes, & Janssen, 1994; Orizio et al., 1999; Vøllestad, Sejersted, & Saugen, 1997). Changes in both EMG and MMG signals have been found at contraction levels of 10% MVC (Blangsted, Sjøgaard, Madeleine, Olsen, & Søgaard, 2005) as well as contraction levels below 40% MVC (Madeleine, Jørgensen, Søgaard, Arendt-Nielsen, & Sjøgaard, 2002; Orizio, Perini, Diemont, & Veicsteinas, 1992).

A method that could potentially be used to investigate the cause of back muscle fatigue is near-infrared spectroscopy (NIRS), which measures oxygenation levels of muscle. In conjunction with EMG and MMG, Yoshitake, Ue, Miyazaki and Moritani (2001) found that the restriction of blood flow is one of the most important factors underlying muscle fatigue. Maikala and Bhambhani (2006) have employed this technique during 30 minutes of seated WBV. They found that sitting without a backrest resulted in decreases in oxygenation and blood volume compared to sitting with a backrest during WBV exposure.

7.2.3 Balance

With respect to balance, Study 2 found similar results as in Study 3. One may question our use of a standing balance, as it is a reflection of whole-body postural control, rather than specific to the lumbar area. Seated postural sway measures during unstable seated balance have been used as surrogate measures of trunk postural control (Cholewicki, Polzhofer, & Radebold, 2000). Recently, Slota, Granata and Madigan (2008) measured the effect of seated WBV on trunk postural control using a wobble chair design adapted from Cholewicki et al. (2000). Subjects were exposed to 30 minutes of seated WBV. They found increases in all of their measures after WBV, implying impairment of spinal stability due to WBV exposure.

There were several methodological differences between the study of Slota et al. (2008) and the current research. The WBV level that their subjects were exposed to (1.15 ms^{-2} rms acceleration) was lower to the frequency-weighted acceleration exposure in study 2 but higher than in study 3. Furthermore, the balance test was able to be performed immediately after the seated exposure because the seat served as the measurement instrument. Given that our subjects had to descend from the vibration simulator and then walk to the force plate, any potential effects could have disappeared. Wilder et al. (1996) showed that walking decreased reflex latencies, however their subjects walked for a longer period (5 min) than the subjects in this research. However, unlike the suspension-type seat that was used in study 3, the seat that they used was not representative of any seat found in a real working environment. Furthermore, unlike the design of our studies, they did not present the WBV and no-WBV (control) conditions in a balanced order. Given that the sitting (without WBV) condition was always performed after the (with) WBV condition, there could have been a learning effect. This measure has been shown to be affected by learning (Van Daele, Huyvaert, Hagman, Duquet, Van Gheluwe & Vaes, 2007).

7.2.4 Reflex response

Our results indicate that the reflex response (both the reflex latency and reflex amplitude ratio) was not affected by WBV. These results are in contrast to those found by others (Li et al., 2008; Wilder et al., 1996) who found that there were effects of WBV. In study 2, the vibration magnitude was 1.4 ms^{-2} rms (dominant frequency around 4.5 Hz), which was higher than those by Li et al. (2008) [0.22 ms^{-2} rms] and Wilder et al. (1996) [0.315 ms^{-2} rms]. Moreover, the duration of exposure for this study was a longer duration. However, significant effects of WBV were still not found. Differences in the way in which the sudden load was applied or the manner in which reflex latencies were calculated could also account for differences in the studies.

It was expected that WBV exposure would have had detrimental effects, thereby leading to increased latencies, therefore decreased stability, and reduced balance. However, the results obtained were contrary to what would have been expected. There was a tendency for the latencies to decrease (i.e., ILIO-L3 in both studies), which would in fact indicate an increase in stability. In addition, the value indices of body sway decreased, indicated improved balance. It is possible that these results could correspond to a stiffer trunk and a more rigid body posture after exposure to prolonged sitting. An increase in passive lumbar stiffness has previously demonstrated after prolonged (2 hours) sitting (Beach, Parkinson, Stothart & Callaghan, 2005). Furthermore, it has been shown that the lower lumbar joints approach total range of motion (similar to deep trunk flexion) in seated postures, thus suggesting increased loading of the passive tissues surrounding this region (Dunk, Kedgley, Jenkyn & Callaghan, 2009). These could potentially explain the results found in this research.

7.2.5 Limitations

One limitation of this current research is that the vibration to which the subjects were exposed was applied only in the vertical (*z-axis*) direction. In real occupational environments, workers are subjected to vibrations acting simultaneously in different directions. Further, inasmuch as we tried to simulate working conditions (i.e., seat, posture, vibration characteristics of a LHD vehicle) in the third study, the results of the study would be generalizable to this specific vehicle. Depending on the type of vehicle, the axis with the largest acceleration level could either be in the horizontal (*x-axis*) direction or fore-aft (*y-axis*) direction. In study 3, subjects were also asked to keep their elbows on the armrests as this is the posture adopted by the workers who were observed in the field. However, not all workers may use the armrests. It is possible that different results would have been obtained if the elbows were not supported by the armrests.

Assuming that there are any effects on the dependent variables that we wanted to measure, it is unknown how long any effects persist. It is possible that any effects, if any, disappeared before the post-test was conducted. There was always a time-lag between the end of exposure to the first trial of the post-test. The subjects had to first dismount the vibration simulator and then walk a short distance to perform the post-test.

7.3 Conclusion

This research used Panjabi's model of spinal stability as the framework to: 1) determine the reliability of different biomechanical measures and 2) study possible injury mechanisms related to seated WBV exposure and low back pain and disorders. There was no significant effect of 60 minutes of seated WBV exposure on any of the measures that investigated the three subsystems that control spinal stability. These findings suggest that the mechanisms that lead to low-back pain and disorders may not be related to deficits in spinal stability. Furthermore, it must be remembered that the measures that were used to measure each of the subsystems demonstrated poor to moderate reliability, and some of them are not specific enough to the lumbar area. Therefore, the results of this present research should be taken with some caution.

It is possible, however, that other aspects relating to spinal stability (i.e., proprioception) not investigated in this research could be affected by WBV. These could be considered in future work related to this subject. Furthermore, considering recent findings in other laboratories (i.e., other research groups), the possibility that WBV effects spinal stability remains. Despite not finding any effects of WBV on the biomechanical measures, this research is the first to document an increase in back muscle activation due to WBV in comparison to a no vibration (control, sitting-only) condition. Moreover, unlike the more recent findings that did find effects of WBV, this research: 1) exposed the subjects to a longer duration (60 minutes) of WBV; 2) used a control (no vibration) condition that was counterbalanced with the vibration condition, thus reducing any bias of learning; and 3) attempted to simulate more real-life working conditions such as the vibration characteristics of a mining vehicle, the suspension-type used in this vehicle, and the postures adopted while working in the vehicle.

7.4 Recommendations for future research

Even though results from this research did not show any significant effects on the biomechanical measures, understanding the possible injury mechanisms related to seated WBV exposure is still important. The field of WBV and musculoskeletal disorders remains a "hot" topic for researchers. This has been demonstrated in a recent special issue "Workplace vibration exposure: Characterization, assessment and ergonomic interventions" in the journal *International Journal of Industrial Ergonomics* (Rakheja & Dong, 2008). Further, as with this research, other research groups are hypothesizing a possible injury mechanism as decreased spinal stability (Li et al., 2008; Slota et al., 2008). Interestingly, these studies showed effects of WBV, suggesting that differences in experimental conditions and/or outcome measures may explain these contradictory findings. Clearly, there is still a possibility that WBV affects spinal stability. Studies that consider different WBV exposures and more sensitive outcome measures need to be conducted to clarify this situation.

Further studies could investigate different WBV characteristics (i.e., RMS amplitude, frequency, duration) in multiple directions as long as the exposure remains within the limits set by the International organization for Standardization. This would be more realistic to working conditions experienced by workers. Using this recommendation, the effects using a measure such as the wobble chair (Cholewicki et al., 2000) should be further investigated as it is a better measure of trunk postural control. This method should also be assessed for its reliability. Finally, measures that are more sensitive to detecting muscle fatigue due to low-level contractions should be used as it could lead to a better understanding to the injury pathway. Therefore, future studies with a rigorous experimental design, while taking into consideration important variables (i.e., vibration in other or in multiple directions), and using measures that could have more potential in detecting the effects (e.g., measures more sensitive to fatigue, balance measures more specific to the lumbar region) could advance the knowledge in this area of study and contribute to a better understanding of the link between WBV and low-back pain and injury.

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APPENDIX A:

CONSENT FORMS AND DECLARATION FROM ETHICS COMMITTEE

PARTICIPANT CONSENT FORM

PROJECT TITLE:
Evaluation of whole-body vibration, seat design & performance, and sitting posture in large mobile equipment (Phase III)

PRINCIPAL INVESTIGATORS:
Tammy Eger (Laurentian University)
André Plamondon, Paul-Émile Boileau (IRSST)

DESCRIPTION OF PROJECT AND OBJECTIVES:
Among physical exposures encountered in working conditions, seated whole body vibration (WBV) has been shown to be related to low-back disorders. Evaluation of the biomechanical responses to prolonged WBV exposure is important in improving our understanding of the role WBV might play in the development of low-back pain. The main objective of this project is to develop objective, non-invasive and reliable measures to quantify the biomechanical responses to WBV exposure. The biomechanical measures that will be evaluated are balance, lumbar stability, and spinal height variation. Phase III of the project consists of three separate studies. You will be involved in Study ____ .

Here are the specific objectives associated with each study and the number of times you will be asked to come to the laboratory. Please take note of the study which applies to you.

- ☐ **Study 1:** This study will verify that the targeted biomechanical measures that are expected to be sensitive to WBV can be handled with confidence in our laboratory. The measures will be evaluated to assess their reliability and to some extent, their sensitivity. This requires two visits to the laboratory.

- ☐ **Study 2:** This study will determine which of our measurement techniques is the most sensitive to WBV. This requires six visits to the laboratory.

- ☐ **Study 3:** The one or two most promising biomechanical measures (determined from studies 1 and 2) will be used to quantify the effect of vibration exposure using a seat typical of a mining vehicle. This requires four visits to the laboratory.

NATURE AND DURATION OF PARTICIPATION:
Each experimental session will last between 2 and 3 hours and will take place on the 11th floor of the IRSST (for Study 1) and the 14th floor of the IRSST (for Studies 2 and 3). The day before and the day of the experiment, we would ask you to refrain from any heavy, physically demanding exercise or work. For the measure of lumbar stability,

electrodes will be placed on different back and abdominal muscles to measure the reflex response via electromyography (EMG). Electrodes will also be placed during the vibration conditions to measure the muscular activity, along with an electrogoniometer and light emitting diodes (LED) on your back to measure trunk posture. Only in the sessions where the reflex response is measured will you be asked to perform maximal contractions of your back muscles. You will also perform sub-maximal contractions of your different trunk muscles.

Here are the specific tasks associated with each study. Please take note of the study which applies to you.

- ❑ **Study 1:** During the experimental session, you will perform each of the measurement techniques in the following order: stability, balance, stability, and stadiometry. Between the different measures, a rest period of 5 minutes will be allocated to allow recovery from any effects from the previous measure.

- ❑ **Study 2:** You will perform one experimental condition during each visit to the laboratory, thus totalling six visits. The order in which you perform each condition will be random. The conditions are as follows: 1) no vibration/stability; 2) vibration/stability; 3) no vibration/balance; 4) vibration/balance; 5) no vibration/stadiometry; and 6) vibration/stadiometry. Each biomechanical measure will be taken before and after 60 minutes of exposure (vibration or no vibration). EMG will be recorded during this time. Finally, in the conditions <u>with</u> vibration, the recovery from the effects of WBV will be evaluated every 20 minutes post-exposure for a period of 1 hour.

- ❑ **Study 3:** You will perform one experimental condition during each visit to the laboratory, thus totalling four visits. The order in which you perform each condition will be random. In each condition, you will be exposed to 60 minutes of vibration while sitting on typical seat of a mining vehicle. The conditions are as follows: 1) asymmetric posture/stability/with vibration ; 2) a asymmetric posture/stability/no vibration; 3) asymmetric posture/balance/with vibration; and 4) asymmetric posture/ balance/no vibration.

RISKS AND TERMINATION OF THE STUDY :

The risk to vibration exposure is minimal since the levels of vibration and corresponding exposure durations will be maintained below the criterion defining safe exposure in the international standard ISO 2631-1:1997 and ISO 13090-1:1998. Skin LED markers, the electrogoniometer, and surface EMG electrodes will pose no risk. Please inform us if you have any skin allergies to rubbing alcohol. Maximal back extension exertions should not harm you. They will be performed isometrically and the trunk will be in a neutral position. Furthermore, 2 minutes of rest will be given between each contraction.

If, over the course of the experiment, you experience any pain or sickness, please advise the experimenter immediately. You are free to withdraw from the study at any time and you are under no obligation to give a reason for your withdrawal or to return for further experimentation. At any time, you may push the "kill switch" to stop the motion simulator.

CONFIDENTIALITY :
The data collected during the experiment will be kept confidential and anonymous. The results will be treated in a confidential manner and will be used exclusively in the framework of this study. No mention of your identity will be made during the publication of the results of this study.

REMUNERATION :
You will be remunerated a maximum of $ __________ for the completion of all necessary conditions.

CONTACT PERSON :
The resource/contact person appointed to this project is Brenda Santos, who may be contacted at the IRSST by telephone (514) 288-1551 ext. 400, by fax (514) 288-6097, or by email santos.brenda@irsst.qc.ca.

CONSENT :
I undersigned, ________________________________, declare to have received all the information concerning the goals and the outline of this present study. I am fully aware of my responsibilities and I accept to volunteer to participate in this study. I also understand that I may withdraw from the study at any time.

A copy of the information and signed consent form has been given to me.

________________________________ ________________________

Signature of participant Date

________________________________ ________________________

Signature of witness Date

Laurentian Laurentienne

Research Ethics Board
School of Graduate Studies and Research
L-335-A
(705) 675-1151, ext 3213
(705) 671-3840
gmiller@laurentian.ca

This is to certify that the research proposal entitled *Evaluation of Whole-body-vibration, seat design and Performance, and Sitting Posture in Large Mobile Equipment, File #2003-06-01*, submitted by Tammy Eger, with Alan Salmoni, André Plamondon and Paul-Emile Boileau **on 06/03/2003 has passed an ethics review by the Laurentian University Research Ethics Board.**

Conditions: N/A

Signed [illegible] **Chairperson of Ethics Committee**

Signatures of Members	Department
[illegible]	Sudbury & Dist. Health Unit
[illegible]	Nursing
[illegible]	Commerce & Administration
[illegible]	Philosophy
[illegible]	Psychology
[illegible]	DEVE

Date: 18 June 2003

Note: this approval covers only the documents submitted, in the language in which they have been submitted. Any changes to questionnaires or procedures must be re-submitted to the Board, as stated on the form.

Start Date: September 2003 Finish Date: September 2005
Report Date(s): End of September 2004, 2005

APPENDIX B:

BAECKE HABITUAL PHYSICAL ACTIVITY QUESTIONNAIRE

# Item	Question	Score (reserved space)					
1.	What is your main occupation? ______________________		1	3	5		
2.	At work, I sit : never / seldom / sometimes / often / always		1	2	3	4	5
3.	At work, I stand : never / seldom / sometimes / often / always		1	2	3	4	5
4.	At work, I work : never / seldom / sometimes / often / always		1	2	3	4	5
5.	At work, I lift heavy loads : never / seldom / sometimes / often / always		1	2	3	4	5
6.	After working I am tired : very often / often / sometimes / seldom / never		5	4	3	2	1
7.	At work I sweat : very often / often / sometimes / seldom / never		5	4	3	2	1
8.	In comparison with others of my own age I think my work is physically : much heavier / heavier / as heavy / easier / much lighter		5	4	3	2	1
9.	Do you play a sport? Yes / No						
	If yes :						
	- which sport do you play most frequently? ______________	Intensity	0.76		1.26		1.76
	- how many hours per week ? < 1 / 1-2 / 2-3 / 3-4 / > 4 hours	Time	0.5	1.5	2.5	3.5	4.5
	- how many months per year ? < 1 / 1-3 / 4-6 / 7-9 / > 9 months	Proportion	0.04	0.17	0.42	0.67	0.92
	If you play a second sport :						
	- which sport is it ? ______________	Intensity	0.76		1.26		1.76
	- how many hours per week ? < 1 / 1-2 / 2-3 / 3-4 / > 4 hours	Time	0.5	1.5	2.5	3.5	4.5
	- how many months per year ? < 1 / 1-3 / 4-6 / 7-9 / > 9 months	Proportion	0.04	0.17	0.42	0.67	0.92
10.	In comparison with others of my own age I think my physical activity during leisure time is : much more / more / the same / less / much less		5	4	3	2	1
11.	During leisure time I sweat : very often / often / sometimes / seldom / never		5	4	3	2	1
12.	During leisure time I play sport : never / seldom / sometimes / often / always		1	2	3	4	5
13.	During leisure time I watch television : never / seldom / sometimes / often / always		1	2	3	4	5
14.	During leisure time I walk : never / seldom / sometimes / often / always		1	2	3	4	5
15.	During leisure time I cycle : never / seldom / sometimes / often / always		1	2	3	4	5
16.	How many minutes do walk and/or cycle per day to and from work, school and shopping? < 5 / 5-15 / 15-30 / 30-45 / > 45 minutes		1	2	3	4	5

Calculation of the simple sport-score (I_9):

$$I_9 = \sum_{i=1}^{2} (\mathrm{int}\,ensit\acute{e} \times temps \times proportion)$$

$$= 0 \;/\; 0.1 - < 4 \;/\; 4 - < 8 \;/\; 8 - < 12 \;/\; \geq 12$$

1 2 3 4 5

Calculation of scores of the indices of physical activity :

Work index = $[I_1 + (6 - I_2) + I_3 + I_4 + I_5 + I_6 + I_7 + I_8] / 8$

Sport index = $[I_9 + I_{10} + I_{11} + I_{12}] / 4$

Leisure-time index = $[(6 - I_{13}) + I_{14} + I_{15} + I_{16}] / 4$

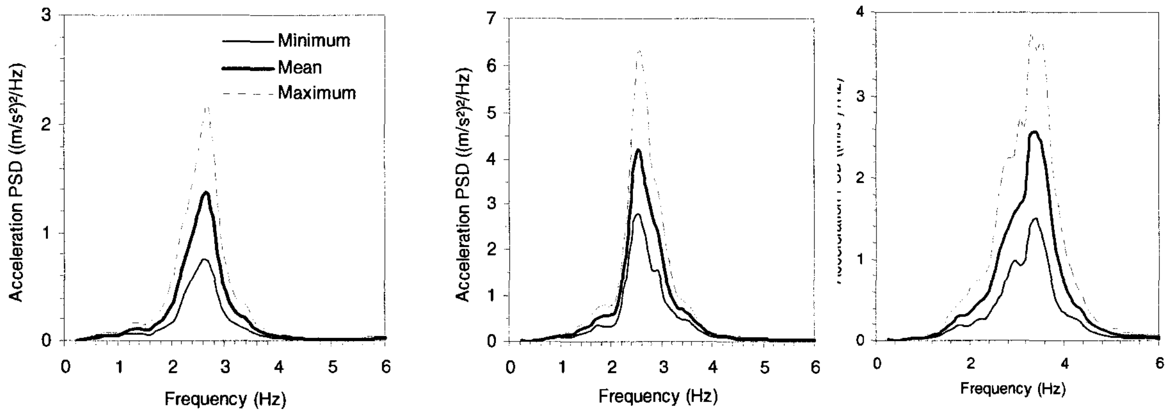
(Baecke, Burema & Frijters, 1992)

APPENDIX C:

SPECTRAL CLASS CHARACTERISTICS

Vibration spectral characteristics of the large and small LHD vehicles tested in an effort to categorize the vehicles in terms of vibration spectral classes to be reproduced on a laboratory whole-body vibration simulator to assess the vibration attenuation performance of a typical LHD suspension seat. Vertical vibration measured at the seat attachment (floor) of 8 small and 8 large LHD vehicles operating underground in typical mining operations under loaded and unloaded conditions was considered as the basis for defining the spectral classes. By regrouping the data collected for each LHD vehicle size and load condition, the overall distribution of acceleration power spectral density (PSD) of measured floor vibration was determined over the 0.5 to 20 Hz frequency range. Mean and envelopes of maximum and minimum values of PSD spectra were computed to define the spectral classes, along with the corresponding values of frequency weighted rms acceleration determined in accordance to the ISO 2631-1 standard. These spectra were further used to calculate the displacements needed to drive a whole-body vibration simulator consisting of a platform supported by two servo-hydraulic actuators having a total stroke of ±100 mm. Finally, the vibration transmissibility characteristics of a typical suspension seat were determined under sine sweep excitation using both a rigid mass load and a human subject having a mass of 62 kg and 85 kg, respectively. The SEAT value, representing the ratio of seat to base frequency-weighted rms acceleration, was further measured under each of the defined LHD vibration spectral classes by loading the seat with an 85 kg subject.

Three spectral classes applicable to both loaded and unloaded conditions were defined in the figure below (Figure C1): one for large and two for small LHDs. The influence of load on frequency-weighted rms acceleration was found to be negligible for large and Class I small LHDs, while a shift of the peak acceleration PSD to lower frequencies was noted for the loaded vehicles. The influence of load was found to be more important for Class II small LHDs. The following table (Table C1) provides a comparison of frequency weighted, a_w, and unweighted, a, accelerations and dominant frequencies for the mean, maximum and minimum spectra associated with the different spectral classes. These were reproduced on a vibration simulator and used to assess the performance of a typical LHD suspension seat. The results obtained suggest that the seat cannot provide attenuation of the vibration at the dominant frequencies of the vehicles which range from 2.6 to 3.4 Hz . The measured SEAT values ranging from 1.25 for large LHDs to 1.35 for Class II small LHDs confirm that the seat is not adapted to these vehicles.



A

C

Figure C1: Vibration spectral classes: A) large LHD; B) small LHD-Class I: C) small LHD-Class II

Table C1: Characteristics of the spectral classes for large and small LHDs

Spectrum	Large LHDs		Small LHDs-Class I		Small LHDs-Class II	
	a	a_w	a	a_w	a	a_w
Minimum (ms^{-2})	0.89	0.62	1.63	1.16	1.38	1.13
Mean (ms^{-2})	1.20	0.85	2.03	1.45	1.88	1.55
Maximum (ms^{-2})	1.52	1.09	2.45	1.76	2.36	1.95
Dominant frequency	2.7 Hz		2.6 Hz		3.4 Hz	