# THE INFLUENCE OF BODY MASS ON POSTURE, PRESSURE DISTRIBUTION AND DISCOMFORT DURING PROLONGED DRIVING

By

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**Abstract**

**Background:** Currently, if traveling the posted speed limit, the typical commuter driver in the Toronto Metropolitan area will travel round trip upwards of 60 minutes a day to work (Heisz and LaRochelle-Cote, 2005). As urban congestion continues to rise, commuting distances and times will progressively increase, placing commuter drivers at increased risk of developing musculoskeletal disorders (Porter and Gyi, 2002; Walsh et al., 1989; Chen et al., 2005; Sakakibara et al., 2006). As urban areas continue to expand, it is believed that a greater percentage of our urban populations will be defined as overweight or obese (Puska et al., 2003). To date the influence of body mass on driver posture, pressure distribution and discomfort during a prolonged driving situation has been left relatively untested. The purpose of this investigation is to determine the influence body mass has on driver posture, pressure distribution and discomfort during a prolonged driving situation.

**Methodology:** Twelve male and 12 female participants, between 167 and 172 cm in stature were used in this investigation. Even numbers of males were assigned to either a light (51.3-57.7 kg), moderate (63.7-69.4 kg), or heavy (82.7-92.0 kg) body mass group. Participants were than placed in a 2 hour in lab driving simulation. During the simulation, lumbar flexion, pelvic angle, joint/segment angles, pressure distribution and discomfort were recorded. A three way mixed general linear model was used to determine if significant ($\alpha = 0.05$) differences in discomfort, posture and/or interface pressure measurements existed over time.

**Results:** Heavy drivers displayed increased total IT pressures and total seat pan/back pressures during driving. When normalizing these total pressures to area, differences in total IT pressure recorded from the seat pan, and total pressure recorded from the seat back were not significantly different ($\alpha = 0.05$) across body mass groups. Due to the lack of seat pan accommodation with respect to surface area, the heavy body mass group's total pressures per unit area for the seat pan was elevated relative to the lighter body mass groups. No differences in two-dimensional joint or segment kinematics and ratings of perceived discomfort were observed between body mass groups or between genders. Gender specific lumbo-pelvic postures and pressure distribution profiles were observed.

**Conclusion:** With appropriate design of the seat pan to accommodate heavy body mass populations with respect to seat pan area, the influence of body mass as a potential risk factor in the development of discomfort would be reduced. With stature and body mass controlled between gender groups, biomechanical differences in both pressure distribution and lumbo-sacral postures were observed between males and females, verifying gender as a risk factor in the development of discomfort during prolonged driving. Recommendations to car seat manufacturers to recognize gender and body mass as important variables in the design of a car seat should be made.

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,Jon

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## 1.0 Introduction

Canadian urban populations have grown 43.4% in the past 30 years (Bourne, 2005), increasing by 5% from the period of 1996 to 2001 (Statistics Canada, 2006). With a lack of rural planning regulation in North America, "out-migration" to suburban regions has been the population response to this steady rise of urban congestion in Canadian Metropolitan Areas (CMA) (Guinness and Bradshaw, 1985; Heisz and LaRochelle-Cote, 2005; Herbert, 1972). Coupled with evolving transportation and technology advancements, it is becoming more economically viable for people to live in suburban regions and drive to economic centers for employment (Guinness and Bradshaw, 1985; Herbert, 1972). Out-migration has become the most consistent North American urban geographical trend of the twentieth century (Guinness and Bradshaw, 1985; Herbert, 1972).

Suburban out-migration has forced 50% of the Canadian work force to commute a distance greater than 5km, with approximately 30% of this population commuting over 25km on a daily basis (Heisz and LaRochelle-Cote, 2005). Between 1996 to 2001, over 200,000 (12% increase) additional people began commuting to work on a daily basis in the Toronto metropolitan area alone (Statistics Canada, 2006). Commuting patterns are also becoming increasingly complex as both manufacturing and service industries expanded into growing suburban areas (Heisz and LaRochelle-Cote, 2005). Already stressed urban infrastructure are being progressively loaded as suburb to suburb commuting patterns emerge (Heisz and LaRochelle-Cote, 2005). Currently, if traveling the posted speed limit, the typical commuter in the Toronto metropolitan area will drive upwards of 60 minutes a day to and from work (Heisz and LaRochelle-Cote, 2005). As urban congestion continues to rise, commuting distances and times will progressively increase, placing commuter drivers at increased risk of developing musculoskeletal disorders (Chen et al., 2005; Porter and Gyi, 2002; Sakakibara et al., 2006; Walsh et al., 1989).

It has been shown that increases in commuting distance will increase a person's probability of developing musculoskeletal disorders (Gyi and Porter, 1998; Porter and Gyi, 2002), negatively affecting employee attendance (Chen et al., 2005; Walsh et al., 1989). Currently, it is estimated that 7% of all lost time compensation claims in Ontario were made by drivers placed in prolonged driving situations on a daily basis complaining of lower back pain (LBP)(WSIB, 2004). As time progresses, it is speculated that injury claims will continue to rise as daily commuting time increase, placing increased focus on understanding the possible risk factors associated with commuter driver discomfort.

To date, research investigating driver/car seat interaction is limited, with even less research devoted to the investigations of driver/car seat interaction during a prolonged driving situation. Many investigations of driver/car seat interaction have used self reported ratings of perceived discomfort as their primary measurement tool (Chen et al., 2005; de Looze et al., 2003; Gyi and Porter, 1998; Porter and Gyi, 2002; Sakakibara et al., 2006; Walsh et al., 1989). This approach has proven useful in determining what seat features facilitate the reduction of a driver's self reported perceptions of discomfort (Kolich, 1999), but are limited in explaining how a seat feature affects driver posture, and how these postures may influence discomfort. Discomfort surveys represent a trial and error approach to car seat testing (Kolich and Taboun, 2004), limiting the application of their results to different seating environments or to use in other fields of discomfort research (de Looze et al., 2003; Reed et al., 1991).

Individual population characteristics such as gender, body mass, and stature may all influence driver discomfort. Research has shown that both stature (Na et al., 2005; Reed et al., 2000) and sex (Coke et al., 2007; Na et al., 2005; Reed et al., 2000) affect a person's posture while driving. Investigations of interface pressure distribution during driving have shown that stature and gender will influence ischial tuberosity (IT) pressures (Coke et al., 2007; Gyi and Porter, 1999; Porter et al., 2003), and total seat pan/back pressures (Na et al., 2005) during driving. It is clear that both stature and gender influence driver posture and pressure distribution, exposing these populations to different discomfort pathways when placed in a driving posture. Quantitative measures such as postural and interface pressure recordings have provided the ability to bridge observed mechanical variables to possible discomfort pathways. This information may provide insight into what populations are predisposed to discomfort during prolonged driving situations. This information may then be used by industry to improve car seat design capable of reducing driver discomfort and risk of musculoskeletal disorder during prolonged driving.

A review of the relevant car seat literature has shown that stature consistently influences driver posture and pressure distribution. Hip angle (Na et al., 2005; Porter and Gyi, 1998; Reed et al., 2000), thigh pressure (Gyi and Porter, 1999), total seat pan/back pressure (Na et al., 2005), and seat back/pan pressure change (Na et al., 2005) have all been shown to be influenced by stature.

The influence of gender on driver posture and pressure distribution has produced inconsistent findings within the car seat literature. For example, the literature indicates that males produce larger IT pressures (Coke et al., 2007), larger hip angles (Coke et al., 2007; Porter and Gyi, 1998), and smaller elbow angles (Coke et al., 2007; Park et al., 2000) relative to females while driving. However, other studies have reported that gender does not influence IT pressure (Kolich and Taboun, 2004), hip angle (Park et al., 2000) or elbow angle (Porter and Gyi, 1998) during driving.

The inconsistency in the published literature with respect to the influence of gender on posture and pressure recordings may be due to the lack of anthropometric control demonstrated by most investigations when blocking for gender. Stature and body mass have generally not been controlled for when blocking male and female test populations. Reed et al, (2000) showed that when stature was controlled for between gender groups by matching absolute stature ( $\pm$ 1 cm) versus population estimates of stature ( $50^{th}$ percentile male and $50^{th}$ percentile female), significant postural differences during driving disappeared (Reed et al., 2000).

Of the few studies that have investigated the influence of body mass on driving posture and pressure distribution, body mass has been reported to have no statistical influence (Gyi and Porter, 1998; Kolich and Taboun, 2004). Body mass, however, has only been tested as an interaction with primary independent variables such as stature and/or gender (Kolich and Taboun, 2004; Park et al., 2000; Porter et al., 2003; Porter and Gyi, 1998; Reed et al., 2000). The influence of body mass on driving posture and pressure distribution is therefore not well understood.

As more people are commuting greater distances for longer durations every day (Heisz and LaRochelle-Cote, 2005), the risk of developing musculoskeletal disorders during driving is elevated (Chen et al., 2005; Porter and Gyi, 2002; Sakakibara et al., 2006; Walsh et al., 1989). As Canadians become more urbanized, it has also been shown that more of our population is at greater risk of being defined as clinically overweight or obese (Based on BMI) (Puska et al., 2003). Added incentive to investigate the influence of body mass as a potential risk factor in the development of driver discomfort during prolonged driving is becoming more apparent as Canadian urban trends persist.

### 1.1.1 **<u>Research Question 1:</u>**

Does body mass and/or gender influence hip angle, trunk angle, pelvic angle, and lumbar flexion and do these measures vary with time during a prolonged driving situation?

### 1.1.2 **<u>Research Question 2:</u>**

Does body mass and/or gender influence static seat pressure distributions and do these seat pressure recording vary over time during prolonged driving?

### 1.1.3 **<u>Research Question 3:</u>**

Does body mass and/or gender affect self-reported ratings of perceived discomfort, and do these ratings of perceived discomfort vary with time during prolonged driving situations?

### 1.2.1 **<u>Statement of Hypothesis #1</u>**

Increases in body mass and females will be associated an increase in hip and trunk angle, the pelvis will be placed into more posterior rotation relative to the vertical axis and the lumbar spine will become more flexed relative to a neutral standing posture.

All of these dependant variables will increase as time progresses.

### 1.2.2 **<u>Statement of Hypothesis #2</u>**

Both the left and right IT total pressures and seat pan/back total pressures will increase as a person's body mass is increases during prolonged driving. Males and females will display different IT and seat back/pan total pressures during prolonged driving.

### 1.2.3 **<u>Statement of Hypothesis #3</u>**

Increases in body mass and females will produce elevated ratings of perceived discomfort in the buttocks and lumbar region. All body regions will display an increase in discomfort over time for all body mass groups.

## 2.0 <u>Literature Review</u>

It has been shown that when a driver's commuting distance/time are increase, their probability of absenteeism from work is increased (Porter and Gyi, 2002). Increased driving mileage and time has also been shown to increase a persons probability of developing musculoskeletal disorder during driving (Chen et al., 2005; de Looze et al., 2003; Gyi and Porter, 1998; Porter and Gyi, 2002; Sakakibara et al., 2006; Walsh et al., 1989). Findings form an investigation conducted by Walsh et al., (1989) showed that when men drove four hours a day or more as a part of their profession, their probability of developing lower back pain was increased by 1.6% relative to populations who did not drive as a part of there profession. These findings were echoed by Chen et al., (2005) who found that when working populations were exposed to prolonged driving situations (>4 hrs a day) they were at an increased risk (1.79%) of developing lower back pain (Walsh et al., 1989). Surprisingly, the development of low back pain in professions that expose workers to prolonged driving situations (> 4 hrs a day) were found to be at greater risk of developing lower back pain relative to men who use jackhammers as a part of their profession (Walsh et al., 1989).

A cross-sectional questionnaire study of a police population (N=171), revealed that 18% of a this population who drove over 25,000 kilometers per year as a part of their job "always" or "often" experienced lower back discomfort during driving, missing on average, 2.84 days of work per year (Gyi and Porter, 1998). When compared to police population that drove less than 15 000 km/yr, only 2% of this population "always" or "often" experienced lower back discomfort, only missing, on average, 0.67 days of work per year (Gyi and Porter, 1998). In a similar study conducted by the same research group of an normal British public population (male N = 303; female N = 297), they found that people who commuted distances greater than 25, 000 km/yr missed on average 22.4 days of work due to low back pain (Porter and Gyi, 2002). People who drove less than 5,000 km to work each year reportedly missed 3.3 day of work as a result of lower back pain. A relationship between increased mileage and increased reports of lower back discomfort during driving was concluded (Porter and Gyi, 2002).

In a recent cross-sectional questionnaire, representative of a heath care population (N=551), it was found that people who experienced lower back pain (LBP) at the time of their survey, drove an average, 2,000 km per week (Sakakibara et al., 2006). People who did not experience lower back pain at the time of survey, drove an average, 1,000 km per week (Sakakibara et al., 2006). The final conclusions of this study corresponded with previous research groups, suggesting that increasing a person's driving mileage, increases their probability of developing LBP. When comparing the LBP pain population's daily commuting distance to the non-LBP population's daily commuting distance and times, they were found to be equivalent (Sakakibara et al., 2006). A secondary conclusion made by this research group was that the development of LBP may also be a function of the number of days a person commutes to work in sequence per week (Sakakibara et al., 2006).

Investigations of a drivers posture during prolonged driving have shown that a driver's pelvis is rotates in a posterior direction and the lumbar spine is forced into spinal flexion when placed in a driving posture (Beach et al, 2007, Coke et al., 2007; Reed et al., 1991). During a prolonged driving situation, driver posture has been classified as static and flexed (Coke et al., 2007; Grieco, 1986; Kolich and Taboun, 2002). Static lumbar flexion postures have been shown to initiated the creep response in the passive tissues of the lower back

(McGill and Brown, 1992). Flexed postures have also been shown to increase a person's probability of degenerating their intervetebral disks (Videman et al., 1990) and development of lumbar disc herniation (Kelsey and Hardy, 1975).  With the passive tissues of the lower back (ligaments, fascia, facet joints etc..) (Jackson et al., 1966; Pedersen et al., 1956) and the outer edges of the intervetebral disks (Cavanaugh, 1995) being innervated with myelinated and unmyelinated nociceptors; loading these tissues with prolonged lumbar flexion, which are adopted during prolonged, may induce an elevated pain response in the lumbar region during a prolonged driving.

Some investigators believe that as drivers are placed in static postures for long periods of time, the muscles of the lower back are forced to maintain static contractions to the moments that support the upper bodies Centre of Mass (CoM) during driving (Reed et al, 1991). Prolonged low level static contractions (<2% MVC) have been shown to decrease muscle oxygenation, (McGill et al., 2000), and increase muscular fatigue (Jorgensen et al., 1988). Decreased muscle oxygenation is thought to facilitate the development localized ischemia, allowing metabolites and blood pH to pool locally, initiating a localized pain response in the lumbar region (Gregory, 2005). It is also thought that as the lower back muscles fatigue as driving time progresses, they also lose their capacity to maintain the upper body's centre of mass in an upright position (Reed et al., 1991), allowing the spine to be placed in a more flexed posture as driving progresses. A more flexed posture would in turn perpetuate the creep response of the passive tissue in the lower back (McGill and Brown, 1992), initiating a pain pathway from these tissues (Cavanaugh, 1995; Jackson et al., 1966; Pedersen et al., 1956). Until mean/median power frequency shifts, characteristic of muscular fatigue are measured during driving representative of muscular fatigue (El et al., 2003; Kolich et al., 2000; Kolich and Taboun, 2004; Kroch-Lund and Jorgensen, 1991), this proposed mechanism of lower back discomfort can not be validated

More recent research suggests that this positive feedback cycle of increased lumbar flexion, muscular fatigue and lower back discomfort does is not supported by previous reports in the office seating literature. The concept of the Flexion Relaxation Phenomenon (FRP) has been brought to the seating literature in efforts to describe a potential link between driving posture and lower back discomfort. It has been shown that myoelectric activity of the thoracic erector spinae (Callaghan and Dunk, 2002; O'Sullivan et al., 2006), and the superficial lumbar multifidus and the internal obliques (O'Sullivan et al., 2006) decreases as the lumbar spine is flexed past approximately 30 to 40 degrees of normalized lumbar flexion (Callaghan and Dunk, 2002; O'Sullivan et al., 2006). When the lumbar spine is flexed past this threshold, FR is induced, placing the upper body moments, previously supported active muscle tissue, to passive tissues (facet joints, ligaments, fascia etc.) (Callaghan and Dunk, 2002; Floyd and Silver, 1955; O'Sullivan et al., 2006). As the passive tissues are loaded for long periods of time, these highly innervated tissues could initiate a pain response (Cavanaugh, 1995; Jackson et al., 1966; Pedersen et al., 1956), increasing lower back discomfort during prolonged driving (Callaghan and Dunk, 2002; O'Sullivan et al., 2006).

Myoelectric measurements of the lower back musculature and passive tissue testing of the lower back need to be conducted before the FRP can be verified during a driving seating environment. Coke et al, (2007), have shown that the male lumbar spine flexes to approximately 40% max lumbar flexion, while females flex to approximately 47% max lumbar flexion during driving (Coke et al., 2007). This corresponds to approximately 30º and 40º of lumbar flexion during driving for males and females respectively (Coke et al., 2007). If the onset of the FRP is similar between an office and driving seating environments, it is possible that the FRP occurs during driving. This suggests that the FRP is the likely mechanism loading the passive tissues of the lower back during driving; a more appropriate mechanism is describing lower back discomfort in a prolonged driving situation.

With only theoretical knowledge of how lower back discomfort develops or progresses during driving, there is even less knowledge pertaining to how stature, gender and body mass may affect the development or progression of lower back discomfort during a prolonged driving situation. Most car seat literature evaluate driver discomfort with short term (60 seconds to 30 minutes) evaluations of car seat/human interactions, which have been shown to be inappropriate time intervals to test driver discomfort (Gyi and Porter, 1999).

Stature has been shown to influence how a person adopts a driving posture and distributes their pressure over the seat pan while driving (Gyi and Porter, 1999; Na et al., 2005; Park et al., 2000; Porter and Gyi, 1998; Reed et al., 2000). Gender as an independent variable has proven inconsistent in producing distinctive joint angles and pressure distributions during driving (Coke et al., 2007; Gyi and Porter, 1999; Porter et al., 2003). The influence of body mass on pressure distribution and posture during driving has not been tested as an independent variable, leaving its influence relatively unknown.

Static pressure distribution and two-dimensional (2D) kinematics have been fundamental tools used to quantitatively measure participant/car seat interactions. Static pressure distribution recordings have been shown to be influenced by stature, gender groups, and body mass (Gyi and Porter, 1999; Porter et al., 2003). 2D kinematics have been capable of recording distinctive relative body angles as influenced by a person's stature (Park et al., 2000; Porter and Gyi, 1998), but have proven inconsistent in repeatedly demonstrating differences between gender groups. The influence of body mass on pressure distribution and posture has been left relatively untested.

The use of dynamic pressure distributions has been shown to produce unique readings with respect to different stature groups (Na et al., 2005), and may prove to be a promising measurement tool used in prediction and description discomfort during prolonged driving. The use of accelerometers to measure pelvic angle and lumbar flexion as compared to hip angle may also provide a more detailed recording of the pelvic and lumbar flexion response during prolonged driving (Coke et al., 2007). Measuring pelvic angle and lumbar flexion over hip angle may provide more descriptive information in explaining possible mechanical mechanisms that may cause driver discomfort or musculoskeletal disorder during prolonged driving.

### 2.1.0 Prolonged driving

Time intervals used to measure the characteristics of a car seat and how they influence driver discomfort range from 60 seconds to 135 minutes. (Coke et al., 2007; Dunk and Callaghan, 2005; Durkin et al., 2006; Grandjean, 1980; Gyi and Porter, 1999; Kolich et al., 2000; Kolich, 2003a; Kolich and Taboun, 2002; Kolich and Taboun, 2004; Park et al., 2000; Porter et al., 2003a; Porter and Gyi, 1998; Porter and Gyi, 2002; Rebiffe, 1969; Reed et al., 1991; Reed et al., 2000; Thomas et al., 1991) In a study conducted by Porter and Read (1992) as reported by Gyi & Porter (1999), a person's initial ranking of a seat's comfort can be inverted following 135 minutes of driving; or more simply stated, a seat ranked 1st out of 4 seats with respect to subjective discomfort after 15 minutes of testing can receive a rank 4th out of 4 seats with respect to perceived discomfort following 135 minutes of continuous testing (Porter and Read, 1992). From these findings, the Vehicle Ergonomics Group (VEG) has recommended that 120 minutes be the minimum time interval used to accurately test a car seat's comfort (Gyi and Porter, 1999). The use of a 120 minute collection interval to adequately record car seat discomfort has been supported by other researchers in the automotive field (Mergl et al., 2005; Porter and Gyi, 1998; Reed et al., 1991), as significant changes in discomfort have been shown to occur at approximately 80 to 110 minutes of driving, depending on the adjustability of their seat (Gyi & Porter, 1998).

### 2.2.0 Potential risk factors influencing discomfort during prolonged discomfort

Currently, understanding the how discomfort initiates and progresses during a prolonged driving situation is not well understood in the car seating literature. Linking potential mechanical injury mechanisms to discomfort is even less understood. Most of the literature in the car seating field has attempted to link quantitative measures such as peak IT pressure (Gyi and Porter, 1999), total seat pressure (Kolich and Taboun, 2004) and kinematic data such as hip and knee angle (Park et al., 2000; Porter and Gyi, 1998) with subjective measures of discomfort to devise potential mechanical mechanisms with little success.

Stature has been shown to influence a subject's kinematic and pressure distribution profiles while driving. This suggests that stature may be a potential risk factor in the development of discomfort during driving. Variability in gender specific kinematic and pressure distribution profiles reported in the literature has left gender as an undefined potential predictor in the development of discomfort during driving. It is relatively unknown how age and body mass influence a driver's kinematics and pressure distributions during driving, for they have not been tested as independent variable in any short or prolonged driving investigations. Thus their influence as a risk factor in the development of discomfort during driving is left relatively unknown.

Though age may be an important variable in the development of discomfort, it is thought that body mass may be a more influential risk factor influencing driver discomfort in a prolonged driving situation, as heavier people would produce larger contact pressures during driving. By developing a better understanding of how body mass effects a person's seat contact pressure and posture during prolonged driving a better understanding of how body mass may influence discomfort may be more thoroughly explained. Unique mechanical changes as influenced by body mass may also give insight to its role as a potential risk factor in the development of musculoskeletal disorder during driving.

### *2.2.1 Effect of stature on driving posture and pressure distribution*

Most car seats do not accommodate the anthropometrics of the entire driving population (Petherick and Porter, 1996). Typically car seats are designed to accommodate populations in the 50$^{th}$ percentile (Kolich and Taboun, 2002). Populations in the 5$^{th}$ and 90$^{th}$ percentiles may then be at greater risk of developing discomfort during driving; for it has been shown that as the ability of a seat to accommodate a driver is directly proportional to the onset and magnitude of a driver discomfort (Gyi and Porter, 1999; Porter and Gyi, 1998). The ability of a driver to adjust their seat to attain an initial level of comfort may be paramount in the development of discomfort during prolonged driving. By forcing a driver to adopt a posture to fit the geometry of the car seat, rather than adjusting the seat to accommodate the comfort needs of the driver, may then place populations outside the 50th percentile at greater risk of developing musculoskeletal disorder during driving (Kolich and Taboun, 2004; Porter and Gyi, 1998; Porter and Gyi, 2002; Thomas et al., 1991).

Most adjustable features on a car seat are made to accommodate differences n stature, as stature has been shown to be the most influential population characteristic influencing driver posture and pressure distribution during driving. In a study of 16 male Korean students, separated evenly into a tall (average height of 177.3 cm) and a short (average height 168.3 cm) stature groups, the shorter stature group displayed an average greater trunk angle than that of their taller counterparts when the lumbar prominence of their car seat was set to 0 cm in the horizontal, but adopted smaller trunk angles when the lumbar prominence was increased to 3 cm in the horizontal (Na et al., 2005). Another study investigating the postural response of 24 males and 19 females during short term driving simulation (time not reported), found that hip angle was negatively correlated to stature (Park et al., 2000). This study implied that subjects with smaller statures preferred driving with greater hip angles than taller subjects (Park et al., 2000). In a study using 2D passive joint markers, 55 subjects were tested in seven seats for 150 minutes in duration. It was found that the taller subjects experienced a more "open" (larger hip angles) hip posture while driving (Porter and Gyi, 1998). It should be noted however that the hip angles a driver adopts during driving across stature ranges consistently fell within the 100º to 120º range (Coke et al., 2007; Grandjean, 1980; Gyi et al., 1998b; Na et al., 2005; Park et al., 2000; Rebiffe, 1969)

It is evident that hip angle was influenced by stature, but failed to produce a consistent trend across studies. The interior geometries and end range adjustability of the seats were not reported for these studies. This may have been an uncontrolled variable influencing the hip angles of a person while driving. For example, the height of the seat pan relative to the ground may have force a smaller stature populations to adopt a more obtuse knee angle and hip angle to reach the accelerator and break pedals during driving. Also if the lumbar support is set to 100% vs. 0%, a smaller person may be more able to position their seat pan Centre of Pressure (CoP) closer to the front of the seat pan relative to a taller person as a shorter person would posses smaller anthropometrics relative to a taller person (Byers, 2002), giving a smaller stature population more relative mobility when choosing a driving posture. The differences in hip angles reported may have also been affected by the number of subjects tested in each study, the ethnicity of their respective subject pools and method of marker placement on anatomical landmarks. It should be noted that even though studies do not report consisted trends associating stature to specific driving postures; stature has been shown to consistently influence driver posture when stature differences are observed between test populations.

The lack of consistency between stature and driver posture across investigations may be attributed to the lack of sensitivity displayed by rigid linked models in recording hip posture while driving. More detailed investigations of spinal flexion and pelvic angle may provide the detail necessary to produce more consistent relationships between stature and driver posture. Also with this added resolution may provide the added information necessary to explain possible mechanical mechanisms that are associated with lower back discomfort during driving.

### 2.2.2 ***Effect of gender on driving posture and pressure distribution***

As shown by anthropometric data bases, relative to their male counterparts, females posses subtle but obvious anthropometric differences in boney geometry (Byers, 2002; Krongman, 1955). Of all the boney structures in the body, the pelvis is the most dissimilar, for it is the portion of the skeleton that most accommodates the birthing process (Byers, 2002; Krongman, 1955). When comparing a 50$^{th}$ percentile male and female population, relatively speaking, females have flatter wider pelvises, with U versus V shaped sub-pubic angles, elongated ischiums and shorter broader sacrums when compared to males (Byers, 2002; Krongman, 1955). These characteristic sex differences give reason for some researchers to believe males and females may distribute their body weight differently during a prolonged driving situation, placing each gender group at unique risks of developing discomfort or musculoskeletal injury during driving.

Gyi and Porter (1999) found that when tall males (193.9 $\pm$ 8.6 cm) and short females (154.3 $\pm$ 2.2 cm) where placed in seats adjusted outside of their pre-determined "comfortable driving posture", males and females experienced discomfort differently. Tall males produced average static pressure distribution under their IT's that correlated to their subjective ratings of discomfort in the "buttocks" region. Shorter females did not produce any correlations between discomfort and the static pressure distribution recordings under the IT (Gyi and Porter, 1999). Males and females were then concluded to produce unique pressure distributions during driving (Gyi and Porter, 1999). The male group used in this investigation was representative of a 95$^{th}$ percentile population, while the female group was representative of a 5$^{th}$ percentile population. With the influence of stature well documented in the literature, differences in stature between gender groups likely influenced these findings, confounding our understanding of how gender influences a person's pressure distribution while driving  (Gyi and Porter, 1999).

In a short duration (≈ 5 minutes) driving simulation conducted by Park et al., (2000), females adopted a driving posture with smaller elbow angles relative to their male counterparts, while showing no significant differences in hip angle while driving. During a 150 minute driving simulation by Gyi and Porter, (1998), males produced larger more "open" hip angles relative to females while driving, but found no significant differences in elbow angle, opposing Park et al., (2000) findings. In a 60 minute driving simulation study conducted by Coke et al., (2006), there were reported gender difference differences in both hip and elbow angles during driving.   These results corroborate and contradict with both Park et al, (2000) and Porter and Gyi et al, (1998) findings, for both groups found gender differences in either elbow or trunk angle, while Coke et al, (2007) found gender differences in both elbow and trunk angles.  These results show that relative postural angles can be quite variable across studies.  It should be noted that the gender based conclusions made from these papers contained stature differences between gender groups, where the influence of stature may have confound these results.  Park et al (2000) contained an 8 cm stature difference, Porter and Gyi (1998), contained a 17.3 cm stature difference, and the Coke et al, (2007) contained a 10 cm difference between gender groups.  It should also be noted that none of these investigations controlled for seat geometry during the investigation, which may have also attributed to the variable findings.

Chaffin, et al (2000) reported gender differences in reaching tasks with respect to shoulder and elbow angles while driving, demonstrating females interact with the interior geometry of there cars (reaching tasks) differently than males.  In this study it was acknowledged by Chaffin, et al (2000) that these gender differences may have been due to the 13 cm stature difference seen observed between gender blocks. Reed, et al (2000) looked closer to these stature trends, showing that when stature was matched within a range of approximately 1 cm between gender groups; previously observed statistically significant ($\alpha = 0.01$) postural differences as influenced by gender became non-significant post matching. Only 8 participants in total were used when matching the male and female participant, thus may have not possessed enough power to adequately determine if males and females of the same stature display different postures while driving, but did show that stature is a dominant variable influencing driver posture during driving (Reed, et al, 2000). An investigation that controls for stature between gender groups with increased degrees of freedom is needed before more confided conclusion with respect to the influence of gender on driver posture can be made.

### *2.2.3 **Effect of body mass on driving posture and pressure distribution***

With limited to no published literature investigating the influence body mass on participant/car seat interactions and postural kinematics, its influence as a risk factor in the development of discomfort is left relatively unknown. Most studies test the influence of body mass as interactions with gender and stature. With few interactions being reported between body mass and stature, researchers have not found it necessary to test body mass as an independent variable (Gyi and Porter, 1998). With body mass not being tested within definable ranges, it is difficult to determine if different body mass populations responded differently with respect to posture and pressure distribution during prolonged driving. Its influence as risk factor in the development of discomfort during prolonged diving is then left relatively unknown.

Investigations by Coke et al, (2007) have shown that males produced statistically greater total seat pan pressures relative to females during driving, while producing non-significant differences in total seat pan contact area ($\alpha = 0.05$). This can be speculated to mean that males produce greater seat pan total pressures per unit area relative to females during driving. With a mean difference of 14.6 kg between gender groups it is then difficult to determine if these differences in seat pan pressure between males and females are solely attributed to gender or to differences in body mass. As seen with most studies investigating gender as a risk factor in the development of discomfort during driving. Like stature, inherit body mass differences between males and females may confound the observed results (Gyi and Porter, 1999; Park et al, 2000; Coke et al, 2000).

Published literature investigating the characteristics of a driver's seat under load has shown that the foam properties of a car seat is influenced by the loads imposed upon them (Wilson and Blair, 1994). Anthropometrics such as a person's body mass may considerably influence the behavior of the foam properties over time, with respect to total and localization deformation of a car seat (Wilson and Blair, 1994). It has been shown that the rate polyethylene deformation is greatest in the first 20 minutes and continues to deform at a decreased rate for up to 8 hours (Wilson and Blair, 1994). The influence a person's body mass may have on the deformation characteristics of a seat and the pressure response of a driver during deformation may produce valuable information for the onset and response of driver discomfort. A more in-depth investigation of body mass as a primary independent variable is needed if its influence on pressure car seat interface pressure distribution and posture can begin to be answered. Dependant variables such as hip angle, lumbar flexion, pelvis angle, and pressure distribution over time may provide the information necessary to clarifying the role body mass may play in the response of driver discomfort. They may also answer questions regarding potential mechanical changes influenced by a person's body mass, giving valuable insight to potential mechanisms associated with musculoskeletal disorder during prolonged driving.

**2.3.0 <u>Self reported measurements of discomfort</u>**

Shackel et al (1969) were among the first groups to evaluate chair seat comfort. It is from pioneer researchers such Shackel, et al (1969) that current methods for analyzing chair design are rooted. The goal of their research was to determine the short and long term chair comfort so to provide consumers guidelines in selecting a chair for the "user". The method used by this research group was among the first to combine quantitative measures with questionnaires to measure comfort. Video recorded observations of subjects during general use, office use and eating were recorded with ciné-cameras. Their goal was to correlate the frequency of subject body movement to increases in recorded discomfort (Shakel et al., 1969). Movements recorded by the ciné-camera did not correlate well with records of subjective discomfort. However, it was found that self reported questionnaires were capable of assessing chair design as it related to participant discomfort (Shakel et al., 1969).

Hip angle, left leg movement, arm movement, and pressure distribution data have been measured alongside self reported questionnaires in attempts to quantitatively measure discomfort. Due to the inability of the literature to agree upon a widely accepted definition of discomfort or comfort (de Looze et al., 2003) limited the ability of researchers to link quantitative and subjective discomfort measures. Slater (1985) defined comfort as a state of physiological, psychosocial and physical harmony between human and environment (de Looze et al., 2003; Slater, 1985). From a design perspective, the definition of comfort is vague at best, making comfort difficult to measure, interpret and apply.

Discomfort and comfort are thought to be independent measures, where the presence of one represents the absence of an the other (Floyd and Roberts, 1958; Hertzberg, 1958; Zahng et al., 1996). From a research perspective, discomfort may be a more appropriate variable to use when investigating car seat/human interaction (de Looze et al., 2003). Discomfort is coupled with feelings of pain, which are assumed to be caused by the design features of the seat; where relative joint angles, spinal flexion, tissue pressure and circulation are some of the physical factors influencing a person's assessments of discomfort (de Looze et al., 2003; Helander and Zhang, 1997). Discomfort is thought to be a more restricted and definable relative to comfort, since the link between discomfort and physical exposure is more direct (de Looze et al., 2003). This is why discomfort is used as the primary subjective measure when investigating how different test populations are accommodated by different seat designs.

. Currently, self reported discomfort questionnaires are the primary measurement tools used to link the propensity of a seat's design to risk of musculoskeletal disorder while driving (Chen et al., 2005; de Looze et al., 2003; Gyi and Porter, 1998; Porter and Gyi, 2002; Walsh et al., 1989). Kolich, (1999) has shown that self reported questionnaires, when appropriately designed, are reliable, valid measurement tools capable of recording car seat discomfort. Discomfort surveys have been shown to be capable of measuring discomfort in both short and prolonged driving situations, which is why they have been used for all types of car seat research (Coke et al., 2007; Dunk and Callaghan, 2005; Durkin et al., 2006; Gyi and Porter, 1999; Kolich et al., 2000; Kolich and Taboun, 2002; Kolich and Taboun, 2004; Park et al., 2000; Porter et al., 2003a; Porter and Gyi, 1998; Porter and Gyi, 2002; Reed et al., 1991; Reed et al., 2000; Thomas et al., 1991)  Evaluating a seat's design characteristics solely with subjective discomfort evaluations represents a trial and error testing approach, which can be extremely time consuming and expensive (Kolich and Taboun, 2004).

The largest criticism associated with self reported measures of discomfort such as self reported questionnaires are associated with the scales use to record discomfort.  Even though 100 mm visual analogue scales (VAS) have been shown to be sensitive, repeatable and reliable measurement tools of the same individual (Smith et al., 2006), they are limited when used to produce reliable recordings across subjects (Briggs and Closs, 1999). Because a VAS is subjective, each person using them will interpret these scales differently. Each person using these scales will assess their own zero point and distribution range over the analogue scale, giving each person their own error term, making comparison over a population difficult at best (Briggs and Closs, 1999).  VAS have also been shown to possess increased error due to some subject's inability to translate a sensory experience into a linear format (Briggs and Closs, 1999).

Studies have attempted to increase the precision and interpretation of discomfort questionnaires with the use of cross modality matching (CMM)(Reed et al., 1991).  The goal was to use modalities that induced pressure, heat, and cutaneous electro-vibration stimuli to provide added quantitative information in efforts to quantify each person's discomfort response.  Increased heat or pressure would inform the experimenter of circulation disruption.  Cutaneous vibration would give the experimenter information about pain production at the muscular level characteristic of fatigue.  Most importantly these modalities could bring scale to the discomfort response experience by the population.  It was found that the increased precision gained by this technique created an exponential increase in calibration time required for applying this method. The potential benefits of CMM were outweighed by added time requirements needed to apply this method, making it useless in a field setting and/or for investigation with large numbers of subjects unrealistic (Reed et al., 1991). Investigations measuring variables such as body posture and pressure distribution are thought of as more suitable method to gain information regarding possible causes of subject discomfort (Kolich and Taboun, 2004).

### 2.4.0 <u>The use of posture and seat contact pressure measures during prolonged driving</u>

Past research investigating posture while driving has primarily used two-dimensional (2D) relative joint angles as determined by a rigid linked segment models to measure driver posture. The use of 2D postural measures has given researchers the ability to determine if stature and sex influence a subject's general driving posture. This information has been used to determine if stature and sex are potential risk factors for the development of discomfort during prolonged driving (Gyi and Porter, 1999; Na et al., 2005; Park et al., 2000; Porter and Gyi, 1998). Kinematic measures have shown that a driver's posture is influenced by the stature and gender (Gyi and Porter, 1999; Na et al., 2005; Park et al., 2000; Porter and Gyi, 1998). This suggests that stature and gender are influential in the development of musculoskeletal disorders during driving. 2D kinematic data however only provides general information of how gender or stature may affect posture, limiting the ability to determine mechanism leading to musculoskeletal disorders over time. Possessing the ability to directly measure pelvic angle and lumbar flexion over time may provide investigators better insight of how variable such as gender, stature and body mass may place a person at an increased risk of discomfort and development of musculoskeletal disorders during prolonged driving.

Static pressure distribution measures such as average peak pressures and pressure areas have been shown to be capable of correlating a person's ratings of perceive discomfort to pressure distribution recordings (de Looze et al., 2003; Mergl et al., 2005). Static pressure distributing measures however are limited in their capacity to record pressure distribution patterns as predicted by a people of different statures, genders, and body masses (Gyi and Porter, 1999; Porter et al., 2003). Though the static pressure distribution profiles of different populations do not correlate to discomfort across all situations, they still provide a means to determine how a person distributes their peak pressure and total pressures over a car seat. With the ability to measure the pressure exposed to different populations while driving provides insight to their potential in the development of ischemic states while driving (Peters and Swain, 1997; Swain, 2005). The use of static pressure distribution are then capable of providing information to what populations may be at increased risk of discomfort during prolonged driving.

### *2.4.1* ***Posture measures during prolonged driving***

With discomfort being coupled with feelings of pain caused by physical factors such as postural constraints (de Looze et al., 2003; Helander and Zhang, 1997), quantitative measures of sitting and driving postures have been extensively documented in the literature. They have been used as descriptive tools to measure possible cause and progression of discomfort (Coke et al., 2007; Dunk and Callaghan, 2005; Kolich and Taboun, 2004; Na et al., 2005; Park et al., 2000; Porter and Gyi, 1998; Reed et al., 1991; Thomas et al., 1991; Vergara and Page, 2002).

With the use of 2D (Porter and Gyi, 1998; Reed et al., 2000; Thomas et al., 1991) and three-dimensional (3D) motion capture systems (Park et al., 2000), researchers have been able to provide relatively accurate estimations of upper and lower limb kinematics as defined by rigid liked segment models. Most researchers agree that relative knee and hip angles must lie in the range of approximately 120º to 130º and 100º to 120º respectively, to allow subjects to acquire a comfortable seating posture (Grandjean, 1980; Park et al., 2000; Porter and Gyi, 1998; Reed et al., 2000; Thomas et al., 1991). This information can be used by car seat manufacturers as rough guidelines when designing the adjustable ranges of a seat's recline, track position, steering wheel height and seat heights for the driver.

A relatively good blueprint of what hip and knee angles drivers adopt during driving has been documented in the literature (Grandjean, 1980; Park et al., 2000; Porter and Gyi, 1998), but a more detailed schematic of what is happening at the pelvic level and spinal levels are left with relatively poor resolution. Measurements of pelvic angle and lumbar flexion may provide more descriptive, detailed information of what is happening at the lower back. This information may then provide the added information necessary to more confidently determine what populations are at an increased risk of developing musculoskeletal disorders during prolonged driving, and what postures induce lower back discomfort.

Of the investigations studying posture while driving, limited information in how the hip and/or the lumbar spine changes over time during driving has been published (Reed et al, 1991; Na, 2005; Thomas, 1991; Kolich & Taboun, 2004; Kolich, 2003; El Falou, 2003; Park, 2000). As proposed by Reed et al (1991), information of how the pelvis rotates and spine flexes during prolonged driving may provide the information necessary to describe how discomfort develops and progresses and/or relieved over time. Reed et al, (1991) attempted to measure lumbar flexion and pelvic angle during sitting with the use of sonic digitizing probes. Spine, sternum and hip positions were digitized during standing. When the subject was seated in a "comfortable" driving posture, lumbar curvatures were approximated relative to the changes of sternum and hip position when seated. This information was limited by the assumption that changes in sternal and pelvic locations are related to lumbar movement, but did illustrate that the lumbar spine may be placed into flexion when adopting a driving posture (Reed et al., 1991).

Possible methods to directly measure lumbar flexion and pelvic angle during sitting and driving are limited. Methods developed by Dunk and Callaghan, (2005) used "fins" affixed to skin over the C7/T1, T12/L1 and sacrum. Active markers attached to the fin located 6 cm from their attachment to the skin. Using simple trigonometry, relatively accurate, direct measures of spinal angles during sitting were recorded. Due to constraints imposed by a seat back during driving, this "fin" marker method could not be used.

The use of flexible goniometers called rachimeters and an accelerometer method have both provided a means for measuring spinal angles with the use of a seat back like in driving (Coke et al., 2007; Vergara and Page, 2002). It is apparent that both methods have limitations. The rachimeter method is quite large and bulky extending down the entire length of the spine (Vergara and Page, 2002), while the accelerometer method uses 2, 3 X 5 cm pieces of metal located on 2 portions of the lower spinal column to record spinal flexion (Coke et al., 2007). The accelerometer method even though less intrusive assumes a rigid link between the sacrum and pelvis, and assumes that body postures during driving are static or quasi-static. Both methods are limited due to inherit skin movement between transducer and spine, while the rachimeter method may be additionally limited in its capability to be fitted to people with body masses in the 90$^{th}$ percentile. Due to the smaller size and the increased freedom of application displayed by the accelerometer method, it may be a more appropriate transducer to measure hip angle and spinal flexion during a prolonged driving situation across a wider range of test populations.

In a one hour driving simulation, Coke et al., (2006) used accelerometers to measure pelvis angle and lumbar flexion during a driving situation. Their focus was to compare pelvic rotation between males and females from standing posture to driving posture. Statistically there were no differences observed between males and females in pelvic angle and lumbar flexion during driving. This research group did however prove that the pelvis rotates in the posterior direction, while the lumbar spine was placed into flexion relative to standing when placed in a driving posture. This provided valuable information in understanding how the lower back changes from standing to driving, empirically proving that the spine does go into flexion during driving. By showing that the lumbar spine flexes to past 40º of normalized lumbar flexion during driving (Coke et al., 2007), provides evidence that the Flexion Relaxation Phenomenon (FRP) may be induced during a prolonged driving situation. Electromyographical and passive tissue testing must still be conducted before these claims can be verified. This study however did show a possible mechanical link between driver posture, a discomfort model and epidemiological findings, giving better insight into how driver discomfort may develop or progress over time.

### *2.4.2 **Seat contact pressure measures during prolonged driving***

Static interface pressure distributions have been extensively investigated in car seating research (Kolich & Taboun, 2004;(Gyi and Porter, 1999; Porter et al., 2003a). Pressure mats provide researchers the capability to measure seat pan and seat back pressure per unit area during driving. This provides valuable information in determining what areas of the seat back or seat pan may be exposed to increased deformation or predisposes particular body regions in developing an ischemic state. As proposed in previous research, when a body region is placed in an ischemic state, the exchange of metabolic waste products is reduced (Gregory, 2005), which could explain why a person experiences localized ratings of perceived discomfort when placed in a seating environment for long periods of time (Gregory, 2005; Kolich and Taboun, 2004).

Past research has used a threshold of 32 mmHg, which is the approximate pressure measurement of a typical capillary under the human nail as the pressure threshold of pressure sore and discomfort development during sitting (Landis, 1931). This thought process has since been revised, showing that ulcer and discomfort development is more complicated than simply peak pressure development during sitting. The health of supporting structures, such as collagen density, muscle mass density, adipose tissue density, intracellular fluid content and other force distributing tissues surrounding the capillary network under pressure may added information in the development of ischemia and regional discomfort during prolonged driving situations (Swain, 2005).

Of the first models used to describe "safe" interface pressure regions over time, rats were subjected to different loads over different times. A logarithmic decay of allowable peak pressures over time was described, with 90 KPa at time 0 and 20 KPa at 2 hrs being deemed a “safe” interface pressure zone (Kosiak, 1961). How these values translated to the health of a human population is unknown, but it has been shown that increases in pressure are associated with decreases in vascular tissue heath. To date an action limit has not been developed for a human populations. It is known that unhealthy elderly population exhibit larger peak pressures vs. healthy elderly populations (Peters and Swain, 1997). It is then thought that a decrease in peak pressures would be most beneficial for populations without health deformable soft tissues.

Since the 1960’s, investigators have been attempting to correlate body movement (motion capture) and discomfort while sitting (Shackel et al., 1969). The reason is associated with how researchers believe that pressure sores and discomfort development is related to how a person moves with respect to the seat pan during driving and not just by minimizing peak pressures (Swain, 2005). A recognized method to limit the progression and/or development of pressure sores and discomfort is by alternating high and low pressure fields, with the low pressure fields able to sufficiently allow blood flow to return the areas of high relative pressures (Swain, 2005). With advancements in pressure technologies over the past 20 years (Gyi and Porter, 1998), the use of pressure sensors as a means of dynamically track a participant’s movement is becoming more of a reality. With discomfort being thought of as a dynamic phenomenon (Porter et al., 2003), measuring a person’s movement while driving may prove to be a useful descriptor of discomfort. In regions of high pressure, where a person may be placed at an increased risk in developing an ischemic state, an increase in movement may be recruited in attempts to increase the exchange of waste metabolites, serving to decrease localized discomfort (Gregory, 2005).

Reed et al., (1991) were among the first investigators to attempt to measure driver movement with pressure sensor data. They used the standard deviations of the mean pressure values from the seat pan and seat back in attempts to better understand the amount of movement a person experiences during a prolonged driving situation (120 minutes) (Reed et al., 1991). The pressure mats used for this investigation had a resolution of 12 cells over the seat back and seat pan, but sill found correlations between self reported discomfort ratings and the standard deviation recordings produced by a participant's average pressure values (Reed et al., 1991). With the standard deviations of a subject's average pressure distribution correlating with a person's self reported measures of discomfort better than the static pressure measurement itself, suggests the movement of a drivers mean pressure region may be a better descriptor of a driver's discomfort.

Na et al, (2005) defined a "pressure change" (body movement) as a 15% change of average total pressure of the seat back and a 5% change of average total pressure of the seat pan. These definitions were developed from pilot tests that correlated observed movements to pressure changes, producing an $R^2$ of 0.67 for the seat back and $R^2$ of 0.64 for the seat pan (Na et al., 2005). Of the six body regions (neck, shoulder, back, lumbar, hip and thigh) measured over the 45 minute driving simulation, ratings of perceived were discomfort increased in all eight regions (Na et al., 2005). The frequency of pressure change also increased in a corresponding fashion (Na et al., 2005). Pressure change for this investigation did prove useful in reporting unique changes with respect to a person's stature. Pressure change or other dynamic measures of pressure may become useful in finding relationships or trends with respect to gender, stature and body mass. The use of dynamic pressure recording could possibly become a more accurate or sensitive method of determining what variables may predispose someone or increase their probability of developing discomfort while driving. With a percentage cutoff used to describe a person's movement threshold across all people, this definition of movement is then limited to the test conditions used for this investigation. A more robust definition of movement, normalized to the participant is needed if dynamic pressure distributions recordings can be used across investigations and participants.

As CMA's continue to expand (Statistics Canada, 2006) commuting distance and times will also continue to increase (Heisz and LaRochelle-Cote, 2005). If current urbanization trends continue along the same path, it is predicted that more of the Canadian, North American populations will be defined as overweight or obese (Puska et al., 2003). With knowledge of these current population trends it is obvious that research investigating the influence of body mass during prolonged driving is needed. Understanding how body mass influences a subject's static pressure distribution, lumbar flexion, pelvic angle, and hip angle may provide better insight to how discomfort is initiated and/or progresses during prolonged driving. By understanding what variables influence discomfort during prolonged driving, car seat manufactures will be more capable of designing a car seat capable of reducing driver discomfort during prolonged driving for the greater majority of our population. This in turn may then help reduce the thousands of compensation claims made in Ontario by commuters drivers complaining of lower back pain every year or hopefully prevent then number of these claims from increasing in the future (Porter and Gyi, 2002; WSIB, 2004).

## 3.0 <u>Methods</u>

### 3.1.0 <u>Participants</u>

Twelve male and 12 female participants between the ages of 18 and 30 years of age were recruited for this investigation. All subjects recruited had no history of lower back pain causing them to miss a day of work or perform their activities of daily living in the last 12 months. All participants possessed a valid driver license at the time of collection, were drivers of a personal automobile and drove on average at least 30 minutes a day, 3 days a week. Subjects were recruited based on body mass, gender and stature. Gender groups were controlled for both stature and body mass in this investigation. The defined stature range for all participants in this investigation was between 167 cm and 172 cm in height. This corresponded to the stature range of a population after pooling the population estimates of 50$^{th}$ percentile males and females. The 10$^{th}$, 50$^{th}$ and 90$^{th}$ percentile male and female population estimates of body mass were also pooled together to define a light, moderate, and heavy body mass group. Equal numbers of males and females were placed in the three defined body mass ranges. Participants in light body mass group ranged from 51.3 and 57.7 kg, participants in moderate body mass group ranged from 63.7 and 69.4 kg and participants in the heavy body mass group ranged from 82.7 and 92 kg.

#### *3.1.1 <u>Development of stature range and body mass groups</u>*

All anthropometric information used to develop the stature ranges and body mass groups were from Frisancho, (1990). To determine the mean population estimate of stature and body mass in this investigation, both male and female normal distribution curves were pooled together. A pooled variance statistical model was used to express a single estimate of variance with respect to the mean, from multiple subsets of data (Cohen, 1996) (Equation 1).

$$S_p^2 = \frac{(N_1 - 1)(S_1^2) + (N_2 - 1)(S_2^2) + \ldots + (N_k - 1)(S_k^2)}{(N_1 - 1) + (N_2 - 1) + \ldots + (N_k - 1)}$$

Where $N_1, N_2, \ldots N_k$ are the sizes of the data subsets.
$S_1^2, S_2^2, \ldots, S_k^2$ are their respective variances.
k is the subset number.

**Equation 1**: Model used to calculate pooled variance (Cohen, 1996).

To determine the mean population estimate of stature for males/females aged 18-30 years old, estimates of a male stature range in the 50$^{th}$ percentile for 18 – 24.9 year old and 25 - 29.9 year old populations were averaged and pooled. (Table 3.1). The same technique was used to determine the population estimates of the three body mass groups between 18-30 years of age. The population estimates of the 10$^{th}$, 50$^{th}$ and 90$^{th}$ body mass percentiles for males and females were averaged and pooled to develop the light, moderate and heavy body mass groups respectively (Table 3.2).

**Table 3.1:** Mean population estimate of stature for 18-30 year old age group, after pooling the population estimates of 18 – 24.9 year old and 25 - 29.9 year olds.

| Gender | Age (yrs) | N | Variance (cm) | Height (cm) |
|---|---|---|---|---|
| Male | 18-24.9 | 1755 | 49.00 | 176.6 |
| Male | 25-29.9 | 1255 | 49.00 | 176.6 |
| Female | 18-24.9 | 2592 | 42.25 | 163.1 |
| Female | 25-29.9 | 1935 | 36.69 | 162.8 |
| | **Age (yrs)** | **N** | **SD (cm)** | **Height (cm)** |
| **Males/Females** | 18-29.9 | 7537 | 6.60 | 169.8 |

**Table 3.2:** Population estimate of the pooled male/female body mass groups (age range 18-30 years).

| Gender | Age (yrs) | N | Variance | Light | Moderate | Heavy |
|---|---|---|---|---|---|---|
| Male | 18-24.9 | 1758 | 179.6 | 59.8 | 71.4 | 91.5 |
| Male | 25-29.9 | 1256 | 213.2 | 61.8 | 76.0 | 95.1 |
| Female | 18-24.9 | 2592 | 163.8 | 48.4 | 58.3 | 76.1 |
| Female | 25-29.9 | 1935 | 201.6 | 49.0 | 59.4 | 81.6 |
| Males | 18-29.9 | 3014 | 193.6 | 60.8 | 73.7 | 93.3 |
| Females | 18-29.9 | 4527 | 180.0 | 48.7 | 58.8 | 78.8 |
| | **Age (yrs)** | **N** | **SD** | **Light** | **Moderate** | **Heavy** |
| **Males/Females** | 18-29.9 | 7541 | 13.6 | 54.8 | 66.3 | 86.1 |

*Note: All mass values are in kilograms for the light , moderate, and heavy body mass groups.*

From the pooled variance model, the mean population estimate of stature was calculated to be 169.8 $\pm$ 6.6 cm. Considering the stature range used by Reed et al, (2000) to remove stature bias between gender groups ($\pm$ 1 cm), the stature range in this investigation was further restricted to attain a stature range of 170 $\pm$ 2.5 cm (Figure 3.1).

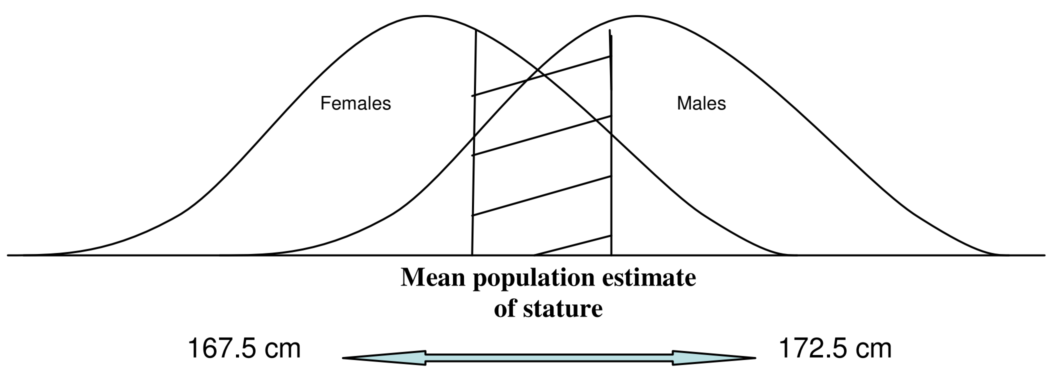


**Figure 3.1**: Mean population estimate of stature (age range 18-30 years).

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The population estimate of a light, moderate, and heavy body mass groups were calculated from the pooled variance model to be 54.75 $\pm$ 13.62 kg, 66.28 $\pm$ 13.62 kg and 86.08 $\pm$ 13.62 kg respectively. To limit body mass ranges from overlapping, a $\pm$ 5 kg within each body mass group was used. The population estimate of a light, moderate, and heavy body mass groups were therefore set to 55 ± 5 kg, 66 ± 5 kg and 86 ± 5 kg respectively (Figure 3.2).

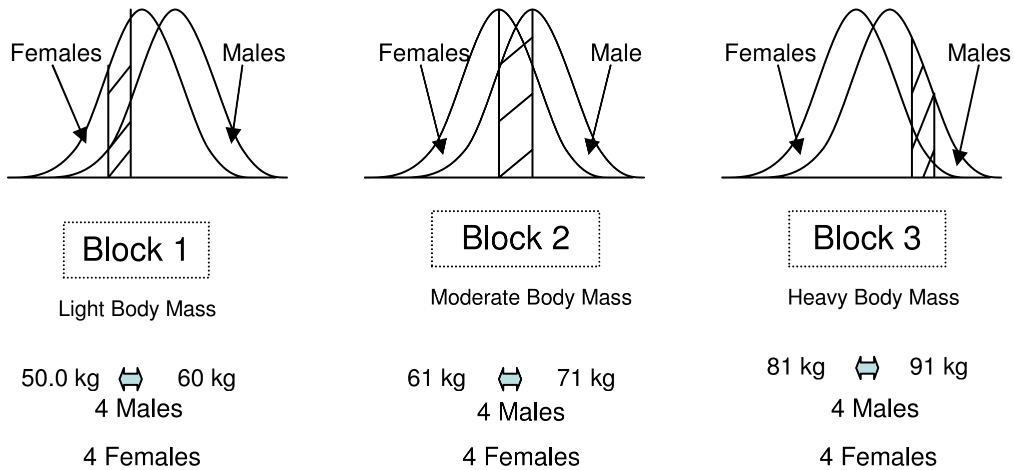



**Figure 3.2**: Population estimate of a light, moderate, and heavy body mass group (age range 18-30 years).

### 3.1.2 *Participant anthropometric information*

Table 3.3 depicts the mean stature, body mass, and girth and age measures of the participants recruited for this investigation when placed in the appropriate body mass groups.

**Table 3.3:** Mean, standard deviation, minimum and maximum values and individual participant body mass, girth and age measures used in the light, moderate, and heavy body mass group.

***Light Body Mass Group***

| | 1M | 2M | 3M | 4M | 1F | 2F | 3F | 4F | **mean** | **SD** | **Min** | **Max** |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| Ht (cm) | 167.0 | 168.0 | 171.0 | 170.0 | 167.0 | 168.0 | 170.5 | 169.5 | 168.69 | *1.79* | ***167.0*** | ***171.0*** |
| Mass (kg) | 57.6 | 51.7 | 57.7 | 54.1 | 56.9 | 54.0 | 51.26 | 56.0 | 54.91 | *2.54* | ***51.3*** | ***57.7*** |
| Girth (cm) | 86.5 | 78.5 | 79.0 | 80.0 | 74.3 | 74.93 | 69.85 | 80.1 | 77.89 | *4.95* | ***69.9*** | ***86.5*** |
| Age (yrs) | 22 | 19 | 19 | 21 | 25 | 25 | 23 | 23 | 22.13 | *2.36* | ***19*** | ***25*** |

***Moderate Mass Group***

| | 5M | 6M | 7M | 8M | 5F | 6F | 7F | 82 | **mean** | **SD** | **Min** | **Max** |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| Ht (cm) | 168.0 | 171.0 | 171.0 | 172.0 | 168.0 | 171.5.0 | 169.0 | 168.0 | 169.81 | *1.73* | ***168.0*** | ***172.0*** |
| Mass (kg) | 67.1 | 69.4 | 63.7 | 65.3 | 64.9 | 68.5 | 64.9 | 64.0 | 65.98 | *2.11* | ***63.7*** | ***69.4*** |
| Girth (cm) | 86.4 | 86.4 | 87.6 | 85.5 | 81.3 | 87.63 | 81.28 | 74.9 | 83.87 | *4.43* | ***74.9*** | ***87.6*** |
| Age (yrs) | 25 | 28 | 20 | 20 | 22 | 24 | 19 | 21 | 22.38 | *3.07* | ***19*** | ***28*** |

***Heavy Body Mass Group***

| | 9M | 10M | 11M | 12M | 9F | 10F | 11F | 12F | **mean** | **SD** | **Min** | **Max** |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| Ht (cm) | 172.0 | 171.0 | 171.0 | 170.0 | 169.0 | 169.5 | 169.0 | 171.0 | 170.33 | *1.08* | ***169.0*** | ***172.0*** |
| Mass (kg) | 84.4 | 84.0 | 83.6 | 82.7 | 84.1 | 83.0 | 92.0 | 90.0 | 83.64 | *0.67* | ***82.7*** | ***92.0*** |
| Girth (cm) | 107.0 | 96.5 | 95.0 | 96.5 | 91.4 | 93.98 | 98.0 | 89.0 | 96.68 | *5.25* | ***89.0*** | ***107.0*** |
| Age (yrs) | 26 | 25 | 21 | 26 | 23 | 27 | 29 | 19 | 24.67 | *2.25* | ***19*** | ***29*** |

*Note: Last letter of subject code (M/F) indicates male or female*

### 3.2.0 Driving simulator

The simulator consisted of a car seat, dashboard, steering wheel, brake pedal, accelerator pedal, and a 22" widescreen computer monitor used as the display system. The car seat used for this investigation was a fully automatic leather truck seat (Ford F1-50). This particular seat model has adjustable horizontal track positioning, seat recline, seat height, pan angle, lumbar support (vertical and anterior/posterior direction) and vertical head rest position.

The interior geometry simulator was adjusted to match that of a mid-sized family sedan (2007 Pontiac G5) (Figure 3.3). All measurements were taken when the seat pan was in full forward position. The left edge of the accelerator pedal was offset to the right of the seats midline by 20 cm. The steering wheel was positioned to be 40 cm from the top of the seat pan to the centre of the steering wheel. The centre of the steering wheel to the lab floor was 75 cm.

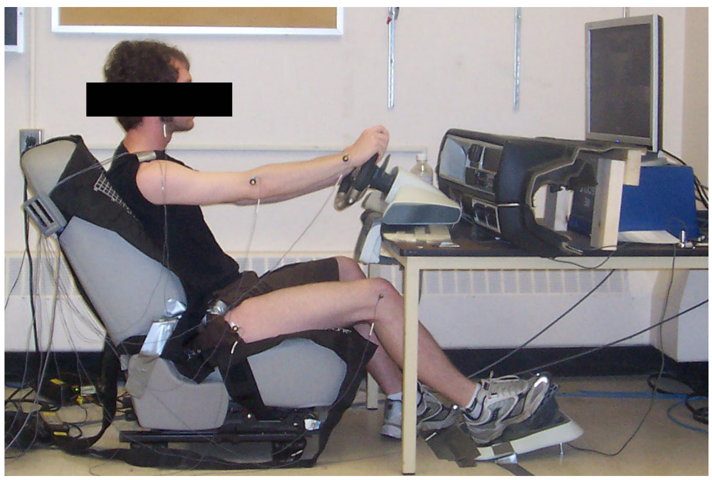


**Figure 3.3:** Depiction of the experimental driving simulator set up.

The seat recline angle was set to 109º, which corresponds to the overlap in hip angles reported by four separate investigations looking at optimal posture and comfort during driving (Grandjean, 1980; Park et al., 2000; Porter and Gyi, 1998; Rebiffe, 1969). Lumbar support was set to 0 in the anterior direction, and left at its lowest vertical point. From the back edge of the set pan to the front edge of the seat pan, the seat pan was inclined 10º degrees relative to the lab space floor. The seat back head rest was removed so the active kinematic marker on the acromium could be tracked with greater accuracy. The only adjustment participants controlled was horizontal track position. Participants in the light, moderate and heavy body mass groups adjusted the seat to a mean distance of 29.4 (4.4), 29.3 (2.5) and 30 (4.2) cm respectively from the front of the seat pan to the base of the accelerator pedal. The horizontal location of the participants hip marker to the base of the accelerator pedal was 66.5 (2.5), 67.2 (2.6) and 67.9 (3.2) cm for the light, moderate and heavy body mass groups respectively.

During driving, all participants were permitted to listen to their choice of music and asked to drive around an oval track staying between 80 and 100 km/h in an effort to simulate highway driving. The simulator was set to an automatic transmission. Video images were created by the commercial gaming software, Project Gotham ™ (Bizarre Creations and Microsoft Corporation, Liverpool, United Kingdom) and run on the X-Box 360 ™ gaming system (Microsoft XBOX technologies Inc., Osaka, Japan).

### **3.3.0 <u>General collection protocol</u>**

#### ***3.3.1 <u>Instrumentation</u>***

Upon arrival each participant was asked to read and sign a University of Waterloo Ethics approved information letter and consent form. Each participant, with shoes off was then measured for stature with a retractable tape measure and for body mass with a standard floor scale (Zenith-products™, Shenzhen, Guangdong, China). A flexible sewing tape measure was used to measure the circumference of each participant's torso in order to approximate girth. The xiphoid process and the inferior boarder of the scapula were the anterior and posterior landmarks used to measure torso circumference.

Each participant then changed into shorts, a T-shirt, and a comfortable pair of shoes appropriate for driving. Participants had both their right sleeve and right short leg rolled and taped to prevent interference with kinematic marker recording during collection.

Participants were then fitted with 2 tri-axial accelerometers (S2-10G-MF, Monitran

Ltd., Penn Bucks, England). One was placed over the first lumbar vertebral body (L1), which was found by counting down from C7, using T3 (level of the scapular spine), T7 (level of the inferior angle of the scapula), and L4 (level with the top of the iliac crest) vertebral locations as reference points (Magee, 1992). The other was placed over the first spinous process of the sacrum (S1) which was found by locating both edges of the posterior superior iliac spine (PSIS), then moving medially to find the second spinous process of the sacrum, which is along the line connecting the two PSIS's (Magee, 1992). The S1 was then found by moving proximal to the level of the S2 (Magee, 1992). The accelerometers were positioned so that the y-axis pointed down and was orthogonal to the ground.

The accelerometers were secured to the skin, using two sided, and medical grade tape. Two, 1 cm thick, 6.5 by 8 cm foam pads were then positioned around each accelerometer (Figure 3.4). The foam pads were used to reduce the discomfort produced by the accelerometers when in contact with the seat back.

A

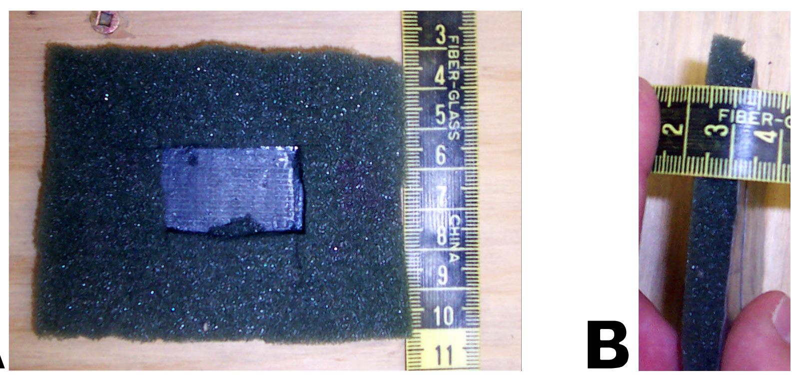



**Figure 3.4:** Frontal (A) and sagittal (B) profile of a typical 6.5 by 8 cm, 1 cm thick foam pads used to decrease the discomfort produced by the tri-axial accelerometers when secured to the spinous process of participants during driving.

Participants were then asked to perform one quiet standing trial with hands in anatomical position to determine each person's "normal" lumbar spinal curvature (Figure 3.5a). Participants then performed three maximum voluntary lumbar flexion trials, where the mean of 3 bending trials was used to determined maximum lumbar flexion (Figure 3.5b) (Coke et al., 2007; Dunk and Callaghan, 2005). After their three bending trials, participants were then fitted with active kinematic markers. Active markers were attached to the targus (skin over top of the ear canal), acromium, lateral epicondyle of the humerus, ulnar styloid, greater trochanter (on skin or overtop of tight fitting clothing), lateral epicondyle of the femur, and lateral malleolus of the fibula (Appendix A). Kinematic markers were only attached to the right side of the participant's body.

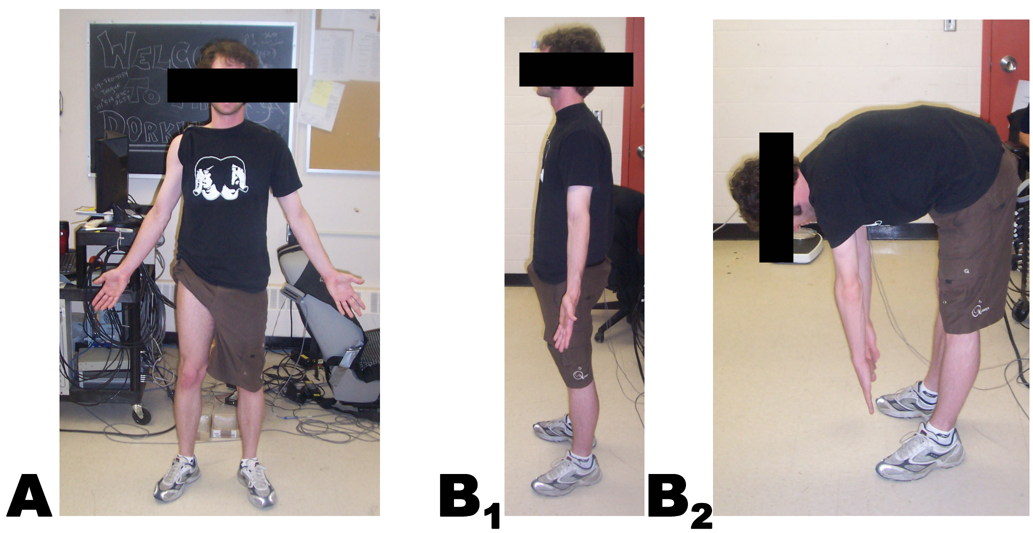



**Figure 3.5:** (A) Frontal view of a participant during quiet standing trial. (B) Sagittal view of a typical quiet stand ($B_1$) and maximum lumbar flexion with legs straight ($B_2$).

After the 3 bending trials, kinematic marker instrumentation and accelerometer instrumentation, participants were asked to sit in the car seat which was fitted with two 36 X 36 sensor pads (X2 Seating System, XSensor Technology Corporation, Calgary, AB) on both the seat back and seat pan (Figure 3.6). A kinematic marker was then placed on a fixed point on the car seat. The lower body kinematic markers were then checked and re-positioned if necessary during this stage in an effort to limit the effects of skin movement that occurred when participants moved from a standing to a driving posture. Once seated, a ten second quiet driving trial was collected so to locate all active markers. Participants then completed discomfort questionnaire 1 and adjusted the track position of the seat to attain a driving posture suitable for driving. The 120 minute driving simulation was then started.

**Figure 3.6:** Depiction of the experimental driver's seat outfitted with two 36 X 36 pressure mapping pads.

### *3.3.2 <u>Data collection</u>*

At time zero participants were asked to fill out a Body Discomfort Survey (BDS) 1, which corresponded to their first contact with the car seat. After BDS 1 was completed and the seat was adjusted to attain a posture suitable for driving the simulation was initiated. Body posture and pressure distribution data were collected in 30 minute intervals starting immediately following the completion of each BDS survey. BDS 2, 3, 4, and 5 were collected at time 30, 60, 90, and 120 minutes respectively. At time 60, seat position was recorded. Both the distance from the front of the seat to the accelerator, and the distance from each participant's greater trochanter (hip) marker to the accelerator were recorded. At time 120, the driving simulation ended. Participants were de-instrumented and given an information package upon exit from lab (Figure 3.7).

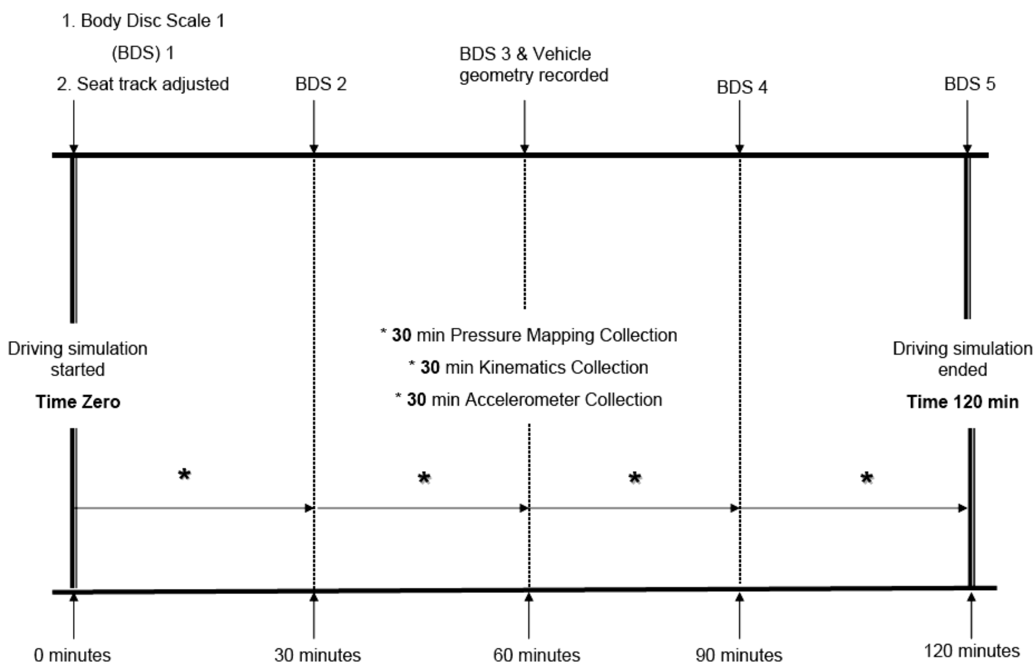



**Figure 3.7:** General experimental time line.

### 3.4.0 <u>Seat pan and seat back pressure distribution recording</u>

Seat pan and seat back interface pressure distributions were recorded on two 36 X 36 sensor pads capable of recording from 10 to 250 mmHg (X2 Seating System, XSensor Technology Corporation, Calgary, AB). Data was sampled at 4 Hz (Dunk and Callaghan, 2005; Coke et al, 2007).

Seat pressure data was recorded from the seat back and seat pan over the 120 minute driving simulation. The pressure distribution data was recorded in 30 minute collection intervals immediately following the completion of the BDS at times 0, 30, 60, 90, 120 (Figure 3.7).

The calibration procedure used to locate the seat pan and pressure mats with respect to the global coordinate system was accomplished by digitizing a point on the seat pan pressure mat and seat back pressure mat with a four-marker rigid body probe (Dunk and Callaghan, 2005). An active kinematic marker was then placed on a fixed point of the car seat. A 10 second quiet driving was trial collected with the seat in the same position that the pressure mats were digitized in. The active marker enabled the absolute location of the seat to be defined. Comparison of the active marker location with the digitized points on the pressure mat enabled the relative location of the seat with respect to the pressure mats to be defined. The calibration routine along with the known dimensions of the pressure mat cells allowed the location of the CoP to be determined with respect to the front of the seat pan (Coke et al., 2007; Dunk and Callaghan, 2005). Due to errors in digitization with the first 9 participants used in this investigation, only 15 of the 24 participants collected in this yielded accurate absolute CoP information with respect to the front of the seat.

Seat pan total pressure (mmHg), seat pan total area ($cm^2$), seat pan pressure per unit area (mmHg/ $cm^2$), left and right IT total pressures (mmHg), left and right total IT areas ($cm^2$), left and right IT pressures per unit area (mmHg/ $cm^2$), seat back total pressure (mmHg), seat back total area ($cm^2$) and seat back pressure per unit area (mmHg/$cm^2$) were all tracked over the 120 minute driving simulation. Total IT pressure and areas were defined as all of the cells adjacent to and including the peak pressure cell that were within 10% of the defined peak pressure value (Dunk and Callaghan, 2005). These were recorded for both the left and right side of the seat pan, so values representative of the right and left IT pressures and areas could be obtained.

### 3.5.0 Body posture/kinematics

For this investigation, two dimensional joint and segment angles, CoP with respect to front of seat pan (m), hip location with respect to the front of the seat pan (m), HAT CoM with respect to the front of the seat pan (m), absolute lumbar flexion values (degrees), normalized lumbar flexion values (% max flexion), pelvic angle with respect to the vertical (degrees) and pelvic angle with respect to standing (degrees) were collected.

To locate the CoP on the seat pan pressure mat in the anterior-posterior direction of the seat pan, each cell's location relative to the front right cell of the seat pan (m) was multiplied with its pressure value (mmHg). The products for each cell were summed, then divided by the total pressure (mmHg) of the seat pan (Winter, 2005). To calculate HAT CoM, the weighted mean position of the hand, forearm, arm, head/neck, and trunk segment centers of mass were determined. The centre of mass calculations included anthropometric data from Winter (1990) and Pearsall et al (1996).

Average relative joint angles were calculated over 30 minute intervals between the initiation and completion of the BDS (Figure 3.7). Using an optoelectronic motion analysis system (Optotrak Centre System, Northern Digital Inc., Waterloo, ON, Canada), the kinematic data was sampled at 32 Hz (Coke et al., 2007), and filtered with a zero-lag fourth-order digital Butterworth low pass filter at 3 Hz (Coke et al., 2007). All data was filtered prior to use in any calculations.

*Hip angle* was defined as the relative angle between a line extending from the greater trochanter through the lateral epicondyle of the femur and a line extending from the greater trochanter to the acromium (Porter and Gyi, 1998).

*Knee angle* was defined as the relative angle between a line extending from the greater trochanter through the lateral epicondyle of the femur and a line extending from lateral malleolus of the fibula to the lateral epicondyle of the femur (Porter and Gyi, 1998).

*Trunk angle* was calculated as the relative angle between a line extending from the greater trochanter to the acromium and a line parallel with the defined horizontal (x) axis of the lab coordinate system (Coke et al, 2007)..

Participant hand placement was not controlled for in this investigation. As a result joint angle measures of the shoulder and elbow were not analyzed or compared to previous investigations in the literature.

### 3.5.1 <u>Pelvic and lumbar spine measures</u>

Accelerometer data was amplified (S2-10G-MF, Monitran Ltd., Penn Bucks, England) to a voltage range of $\pm$ 1 V during calibration, then synchronized with other data sources. Accelerometer data was sampled at 96 Hz (Coke et al., 2007) using a 16-bit A/D conversion system (Optotrak Data Acquisition Unit, Northern Digital Inc., Waterloo, ON, Canada). Accelerometers were calibrated by aligning their sensitive axis at known orientations in the gravitational field (Bouten et al., 1997; Coke et al., 2007; Elble, 2005). All accelerometer data was filtered with a zero-lag fourth-order digital Butterworth low pass filter at 3 Hz (Coke et al., 2007).

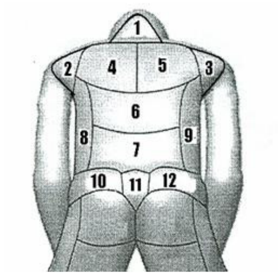
With the assumption of sacro-iliac joint rigidity and negligible inertial accelerations during sitting, inclination of the sacrum relative to its position during upright standing (neutral lumbar spinal position) and rotations relative to the vertical axis (vertical) were considered to be pelvic rotations (Coke et al., 2007). Lumbar spinal flexion was calculated by finding the difference between the inclinations of the accelerometers placed on the S1 and L1. The mean of three maximum voluntary lumbar flexion trials (toe-touching from an upright standing position) recorded prior to active marker were used to normalize lumbar flexion during driving (Coke et al., 2007; Dunk and Callaghan, 2005).

### 3.6.0 Self reported ratings of perceived discomfort measures

A Body Discomfort Scale (BDS) was used to record areas of bodily discomfort (Hartung, 2005; Kolich, 1999; Mergl et al., 2005; Porter and Gyi, 2002; Shackel et al., 1969). A two-dimensional posterior view of a human body, separated into upper and lower body representations of sitting with 20 identifiable body regions was used to locate defined regions of discomfort (Hartung, 2005; Mergl et al., 2005). This survey used 100 mm visual analogue scales to record the ratings of perceived discomfort for the 20 defined regions (de Looze et al., 2003; Smith et al., 2006). Zero mm represented “no discomfort” and 100 mm represented “extreme discomfort” (worst discomfort imaginable) (Figure 3.8). Summed body discomfort was calculated by summing all the ratings of perceived discomfort from the 20 body regions in an attempt to approximate whole body discomfort. The maximum possible rating of perceived discomfort for summed bodily discomfort is 2000 millimeters.

## Body Discomfort Scale

*To answer each question place a slash [ | ] **through** the corresponding line.*



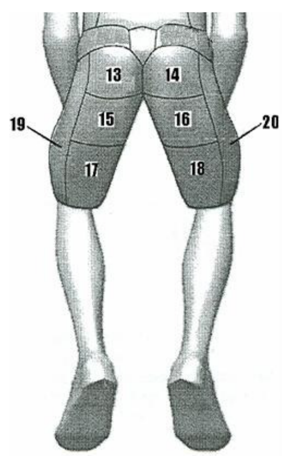

*The number displayed in the regions in the diagram above correspond with the numbers in the survey to the right of the diagram.*

| | No Discomfort | Extreme Discomfort |
|---|---|---|
| 1. Neck | | |
| 2. (L) Shoulder | | |
| R) Shoulder | | 3. |
| 4. (L) Upper Back | | |
| 5. (R) Upper Back | | |
| 6. Middle Back | | |
| 7. Lower Back | | |
| 8. (L) Side of Body | | |
| 9. (R) Side of Body | | |
| 10. (L) Upper Pelvis | | |
| 11. Sacrum/tail bone | | |
| 12. (R) Upper Pelvis | | |

**Scale Continued on NEXT PAGE**

## Body Discomfort Scale (Continued)

*To answer each question place a slash [ | ] **through** the corresponding line.*




| | No Discomfort | Extreme Discomfort |
|---|---|---|
| 13. (L) Buttocks | | |
| 14. (R) Buttocks | | |
| 15. (L) Upper Thigh | | |
| 16. (R) Upper Thigh | | |
| 17. (L) Lower Thigh | | |
| 18. (R) Lower Thigh | | |
| 19. (L) Side of Leg | | |
| 20. (R) Side of Leg | | |

**Figure 3.8:** Body Discomfort Survey (BDS). Manikin used for this investigation is from Hartung, et al, (2005) and Mergl et al, (2005).

### 3.8.0 Statistical Analysis

A three-way mixed general linear model, with gender and body mass as between factors and time as a within factor was used to determine if significant differences in pressure distribution, kinematic and/or discomfort existed ($\alpha = 0.05$). A protected Tukey Honestly Significant Difference (HSD) post hoc ($\alpha = 0.05$) was ran when significant differences were seen between independent variables. When significant interactions with time were encountered ($p < 0.05$), each factor was run as a one way general linear model ANOVA. A protected Tukey HSD was used to determine if significant differences between time intervals existed ($\alpha = 0.05$) (Vincent, 2005).

The stature, body mass, and girth measures within each main effect (gender and body mass) were compared using a one factor ANOVA ($\alpha = 0.05$). A Tukey HSD was used to compare groups if significant differences in stature or body mass were found ($\alpha = 0.05$). This was done to determine if the respective groups were of the same body mass and/or stature. Girth measures were tested with a two-way ANOVA, comparing gender and body mass. A Tukey HSD was used when a significant differences in girth was found between body mass and gender groups ($\alpha = 0.05$). With the use of a one-way ANOVA, girth was tested between genders, within each body mass group. A Tukey HSD was used when significant differences in girth was found within each body mass group ($\alpha = 0.05$). A two-way ANOVA was used to determine if the relative position of the seat and a person's hip marker were adjusted to the same position across gender and body mass groups. A Tukey HSD was used to compare groups if significant differences between independent variables were found ($\alpha = 0.05$).

## 4.0 <u>Results</u>

### 4.1.0 <u>Participant anthropometrics</u>

Body mass groups were not statistically different from each other with respect to stature but were statistically different from each other with respect to body mass ($p = <0.0001$). A Tukey HSD post hoc analysis revealed that all three body mass groups were statistically different from each other. There were no statistical differences with respect to stature and body mass to report when gender was compared.

Girth measures were statistically different between body mass groups ($p < 0.0001$) and between gender groups ($p = 0.0036$). A Tukey HSD post hoc analysis revealed that only the heavy body mass group displayed larger girth values relative to the moderate and light body mass groups. The moderate and light body mass groups were not significantly different from each other. A Tukey post hoc revealed that females were of smaller girth relative to males. When comparing girth measures between males and females within each body mass group, no statistical differences were observed between males and females.

Statistically, the position of the accelerator pedal relative to the front of the seat and the participant's hip marker were not significantly different between body mass groups or gender groups.

## 4.2.0 Static interface pressure recordings

### *4.2.1 Body mass effects*

All differences between body mass groups are recorded in Table 4.1. The values in Table 4.1 represent mean pressure recordings taken over the entire 120 minute driving simulation. Pressures per unit area (PPA) ($mmHg/cm^2$) were used as a means to normalize the areas in which pressure was distributed (Dunk and Callaghan, 2005).

**Table 4.1:** Mean pressure distribution data, grouped according to body mass. All data is expressed as the mean values within each body mass group across all time intervals for the 120 minute driving simulation.

| Measurement Location | Interface Pressure Measurement | Light | | Moderate | | Heavy | |
|---|---|---|---|---|---|---|---|
| | | *Mean* | *(± SD)* | *Mean* | *(± SD)* | *Mean* | *(± SD)* |
| ***Seat Pan*** | Total pressure (mmHg) | 27,358 | (4,072)[xxx] | 30,506 | (1,376)[xxx] | 44,493 | (588)[xxx] |
| | Total area ($cm^2$) | 1,013 | (85)[†,a] | 1,154 | (140)[†,b] | 1,397 | (116)[†,c] |
| | Total PPA ($mmHg/cm^2$) | 27.1 | (4.3)[xxx] | 26.8 | (4.1)[xxx] | 31.9 | (4.2)[xxx] |
| | L IT total pressure (mmHg) | 1,115 | (898) | 1,080 | (1,108) | 2,153 | (1,366) |
| | R IT total pressure (mmHg) | 934 | (884) | 746 | (759) | 1,224 | (1,152) |
| | L IT area ($cm^2$) | 10.6 | (6.8) | 10.6 | (8.6) | 18.9 | (10.4) |
| | R IT area ($cm^2$) | 9.5 | (6.6) | 8.5 | (5.5) | 11.6 | (8.9) |
| | L IT PPA ($mmHg/cm^2$) | 97.3 | (17.8) | 88.6 | (21.9) | 107.9 | (19.2) |
| | R IT PPA ($mmHg/cm^2$) | 85.4 | (27.4) | 74.2 | (27.2) | 97.8 | (16.3) |
| ***Seat Back*** | Total pressure (mmHg) | 6495 | (1,963)[†,a] | 8512 | (795)[†,b] | 10681 | (1,694)[†,c] |
| | Total area($cm^2$) | 479.3 | (115.0)[†,a] | 587.1 | (53.3)[†,b] | 727.9 | (88.0)[†,c] |
| | Total PPA ($mmHg/cm^2$) | 13.5 | (1.6) | 14.5 | (1.2) | 14.7 | (1.2) |

[†] ~ Significant $p < 0.001$; [a,b,c] Tukey post hoc ($p = 0.05$). [xxx] ~ interaction with time.

Significant differences with respect to body mass were found with total seat pan area, seat back total pressure and seat back total area ($\alpha = 0.05$). Significant interactions between body mass and time were observed with respect to seat pan total pressure and seat pan pressure per unit area. All dependent variables showed a significant increase in pressure recordings over time ($p < 0.0001$).

The mean seat pan total area values over the 120 min driving simulation showed significant main effects with respect to mass ($p < 0.0001$). With increased body mass there was an increase in seat pan total area. A Tukey post hoc analysis ($\alpha = 0.05$) revealed that all three groups were significantly different from each other across all intervals over the 120 minute driving simulation (Figure 4.1a).

Seat pan total pressure values displayed an interaction with time during the 120 min collection period ($p < 0.0001$). After conducting a one way GLM on seat pan total pressure, the light, moderate and heavy body mass groups all displayed significant increases in seat pan total pressure over time ($p < 0.001$). Increased body mass was associated with an increase in seat pan total pressure over the entire 120 minute collection. All three groups produced similar increases in seat pan total pressure over time for the first 60 minutes of driving. After 90 minutes, seat pan total pressure was significantly different than the seat pan total pressure at 60 minutes for the moderate and heavy body mass groups. Significant differences in seat pan total pressure relative to the 60 minute time interval did not occur until 105 minutes for the light body mass group (Figure 4.1b). Seat pan total pressures were also elevated with an increase in body mass.

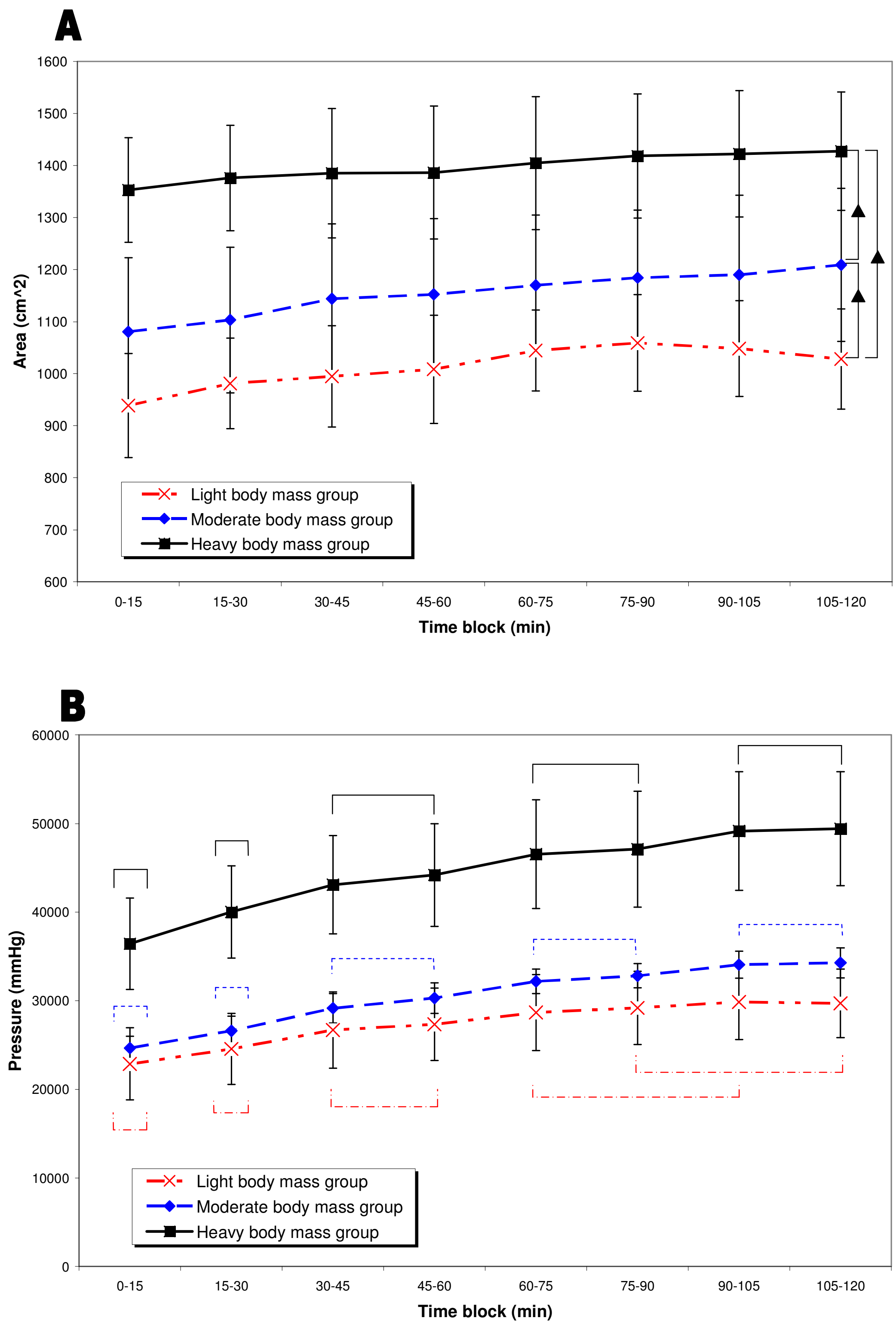



**Figure 4.1:** Time varying response of seat pan total area (A) and seat pan total pressure (B) for the light, moderate, and heavy body mass groups during a 120 minute driving simulation. Each variable is averaged across all participants at 15 minute intervals. ▲~ Tukey post hoc significant difference across all time intervals (α = 0.05). Values grouped under the same horizontal bars in B are not significantly different from each other (α = 0.05).

An interaction between seat pan total PPA and time was observed ($p = 0.0031$). When conducting a one way GLM on seat pan total PPA, the light, moderate, and heavy body mass groups all displayed significant increases over time ($p < 0.0001$). The greatest changes with respect to time were seen in the first 45 minutes for all body mass groups. An increase in body mass was associated with an increased rate of seat pan total PPA change for the entire collection. The seat pan total PPA was similar between the light and average body mass groups over the 120 minute collection. The heavy body mass group displayed increased seat pan total PPA values relative to the lighter body mass groups over the entire collection The seat pan total PPA in the first 15 minutes for the heavy body mass group was 26.9 (3.4) mmHg/cm$^2$. After 15, 30, 60 and 105 minutes relative to the first 15 minute time interval, seat pan total PPA values significantly increased to 29.1 (3.5), 31.2 (4.1), 33.3 (4.4) and 34.7 (4.7) mmHg/cm$^2$ respectively. The seat pan total PPA in the first 15 minutes for the moderate body mass group was 23.2 (3.5) mmHg/cm$^2$. After 30, 60 and 90 minutes relative to the first 15 minute time interval, seat pan total PPA significantly increased to 25.9 (4.0), 27.9 (3.7) and 29.1 (4.9) mmHg/cm$^2$ respectively. The seat pan total PPA of the light body mass group in the first 15 minutes was 24.4 (4.0) mmHg/cm$^2$. After 30 and 105 minutes relative to the first 15 minute time interval, seat pan total PPA increased to 27.0 (4.3) and 29.2 (5.7) mmHg/cm$^2$ respectively (Figure 4.2).

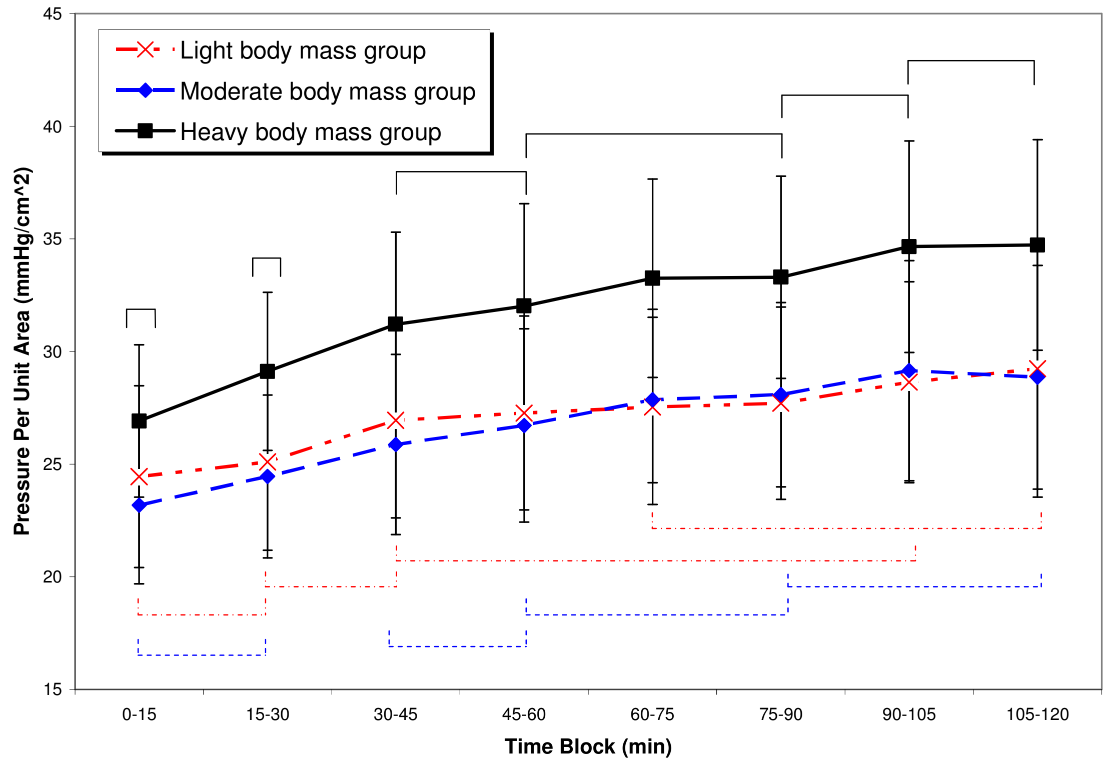



**Figure 4.2:** Time varying response of seat pan total PPA for the light, moderate, and heavy body mass groups. The mean value of each interval is taken across all participants for each 15 minute time interval. Values grouped under the same horizontal bars are not significantly different from each other ($\alpha = 0.05$).

Body mass groups differed with respect to seat back total area ($p < 0.0001$) increasing with body mass group. Higher body mass groups also displayed an increase in seat back total pressure ($p < 0.0001$). A Tukey post hoc showed that all three body mass groups were significantly different from each other at each time interval over the 120 driving simulation for both seat back total area and seat back total pressure.

The seat pan total PPA under the left and right IT and seat back total PPA were not significantly different across body mass groups, indicating that when normalized, the interface pressure profiles display similar pressure outputs (Table 4.1).

### *4.2.2 Gender effects*

All gender differences in seat pressure results are reported in Table 4.2. The values reported in Table 4.2 represent the mean interface pressure recordings taken over the entire 120 minute driving simulation. Pressures per unit area (PPA) (mmHg/cm$^2$) were used as a means to normalize the areas in which pressure was distributed (Dunk and Callaghan, 2005).

**Table 4.2:** Mean pressure distribution data as grouped with respect to gender. All data is expressed as the mean values within each gender group, across all time intervals for the entire 120 minute driving simulation.

| Measurement Location | Interface Pressure Measurement | Males | | Females | |
|---|---|---|---|---|---|
| | | *Mean* | *(± SD)* | *Mean* | *(± SD)* |
| ***Seat Pan*** | Total pressure (mmHg) | 33,566 | (7,693) | 34,672 | (9,745) |
| | Total Area (cm$^2$) | 1,131 | (185) † | 1,245 | (198) † |
| | Total PPA (mmHg/ cm$^2$) | 29.7 | (5.0) | 27.5 | (4.2) |
| | L IT total pressure (mmHg) | 1,370 | (1,332) | 1,528 | (1,111) |
| | R IT total pressure (mmHg) | 1,346 | (1162) | 590 | (361) |
| | L IT area (cm$^2$) | 12.9 | (10.3) | 13.5 | (8.5) |
| | R IT area (cm$^2$) | 12.7 | (8.8) | 7.1 | (8.4) |
| | L IT PPA (mmHg/ cm$^2$) | 95.2 | (21.2) | 100.7 | (20.2) |
| | R IT PPA (mmHg/ cm$^2$) | 94.5 | (23.0) † | 77.1 | (23.1) † |
| ***Seat Back*** | Total pressure (mmHg) | 9,287 | (1,412) † | 7,838 | (2,813) † |
| | Total area (cm$^2$) | 637 | (85) † | 560 | (165) † |
| | Total PPA (mmHg/ cm$^2$) | 14.6 | (1.0) | 13.9 | (1.7) |

† *= significant difference at $p < 0.05$.*

Significant differences with respect to gender were found in total seat pan area, right IT PPA, seat back total pressure and seat back total area. All dependent variables showed a significant increase in pressure recordings over the collection ($p < 0.001$). There were no statistical interactions between time or body mass.

Seat pan total area values were significantly different between genders ($p = 0.0157$), with males covering less of the seat pan area (1,131 (185) cm$^2$) relative to females (1,245 (198) cm$^2$). Gender differences in right IT PP unit area ($p = 0.0487$) showed that males produced larger mean right IT PP unit area (94.5 (23.0) mmHg/cm$^2$) relative to females (77.1 (23.1) mmHg/cm$^2$). Gender differences were seen in seat back total pressure and seat back total area over time, ($p = 0.0102$ and $p = 0.0216$ respectively). Male seat back total pressures were larger (9,287 (1,412) mmHg) relative to females (7,838 (2,813) mmHg). Males displayed a larger seat back total area (637 (85) cm$^2$) relative to females (560 (165) cm$^2$). When the seat pan and seat back data was normalized with respect to pressure per unit area, no significant gender differences were observed (Table 4.1).

### **4.3.0 <u>Kinematics</u>**

Two dimensional joint angles, segment angles, HAT CoM, hip and CoP locations, absolute lumbar flexion angles, normalized lumbar flexion angles (% max flexion), pelvic angle with respect to the vertical axis and pelvic angle with respect to standing were all calculated over the 120 minute driving simulation. Of these postural measures, only pelvic angle with respect to the vertical axis showed significant main effects with both gender and body mass. Normalized lumbar flexion values displayed a 3 way interaction between body mass, gender and time ($p = 0.0397$).

#### ***4.3.1 <u>Gender response: pelvic and lumbar angle</u>***

The pelvis angle with respect to the vertical axis was significantly different between males and females ($p = 0.0151$). Males displayed a more posterior pelvic rotation relative to females, with a mean posterior pelvic rotation angle of 27.9 (6.2) and 22.3 (5.3) degrees, respectively over the 120 min driving simulation (Figure 4.3). No significant changes with respect to time or any interactions were found. No statistical differences or interactions were found for pelvis angle relative to “neutral” upright standing posture, or for absolute lumbar flexion angle.

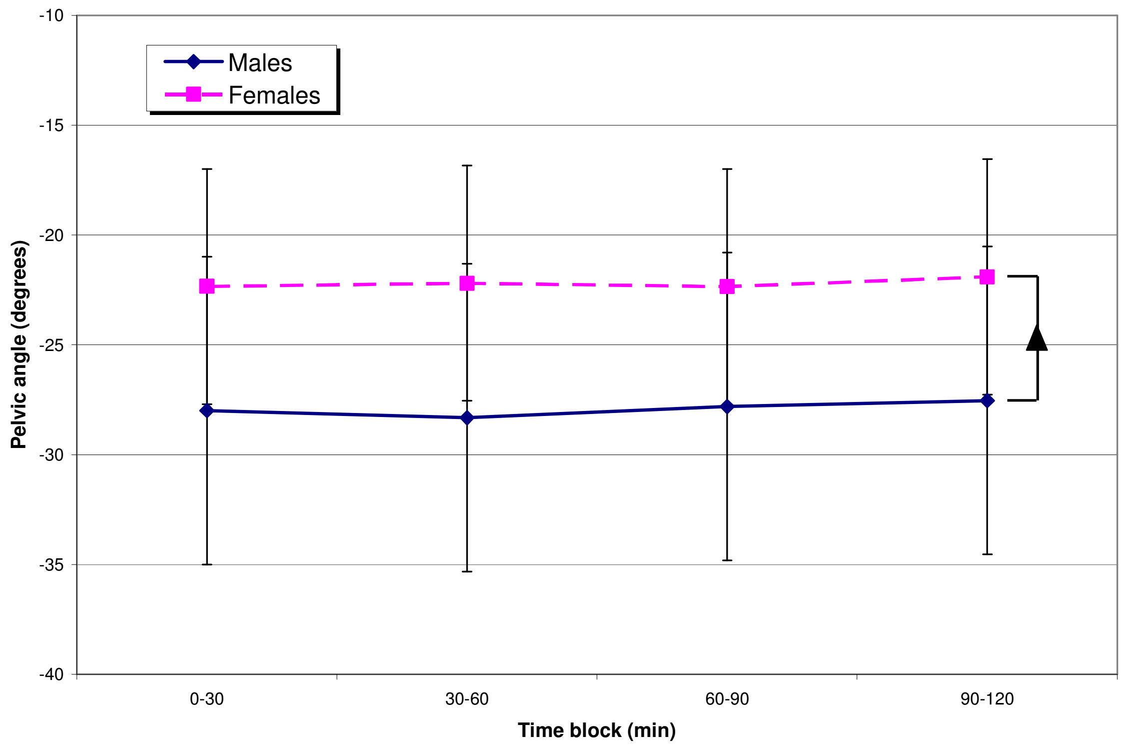



**Figure 4.3:** Time varying responses of pelvic angle with respect to the vertical axis of the lab space as influenced by gender during a 120 minute driving simulation. ▲ ~ signifies a Tukey post hoc significant across all time intervals (α = 0.05).

### *4.3.2 <u>Body mass response: Pelvic and Lumbar Spine</u>*

Pelvic angle with respect to the vertical axis displayed significant main effects for body mass ($p = 0.0397$). Participants in the light, moderate and heavy body mass groups displayed an average posterior pelvic rotation of 22.9 (8.4), 29.2 (5.8) and 23.1 (2.7) degrees respectively over the 120 minute driving simulation. A Tukey HSD post hoc reported no statistical differences between mass groups (Figure 4.4). There were no significant changes with respect to time or interactions to report. No statistical differences or interactions with respect to pelvic angle relative to an upright standing posture were observed. No statistical differences or interactions with absolute lumbar flexion angle were observed in this investigation.

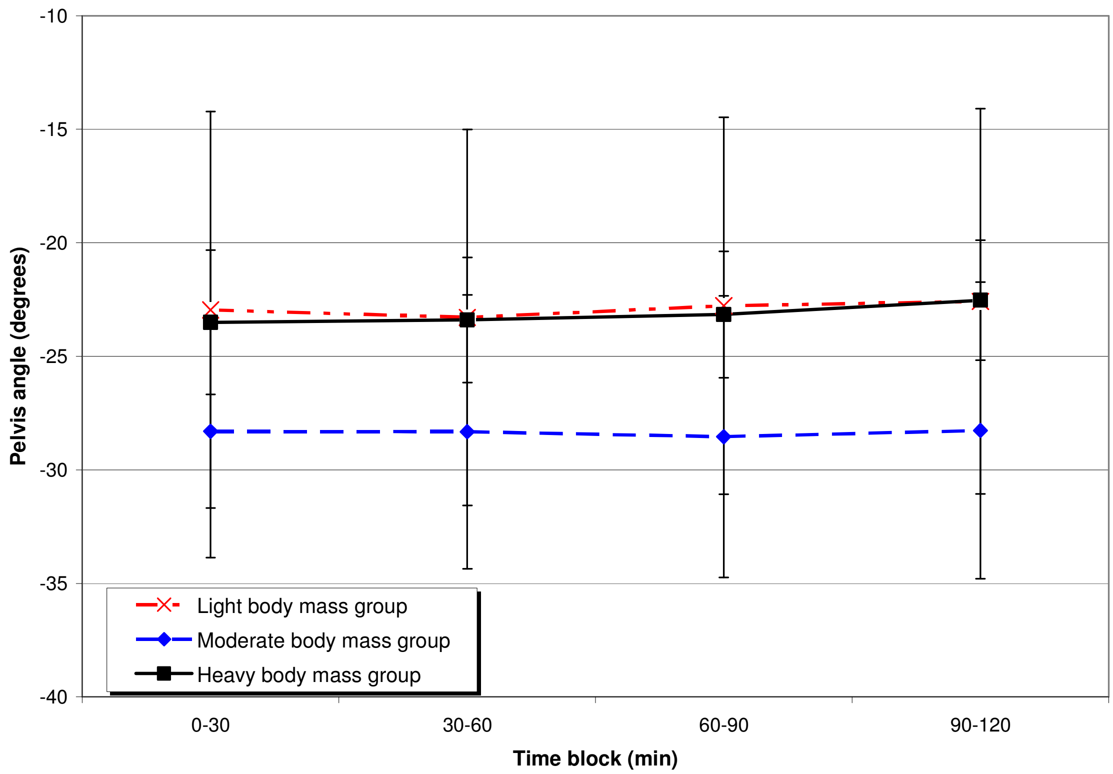



**Figure 4.4:** Time varying responses of pelvic angle with respect to the vertical axis as influenced by body mass during a 120 minute driving simulation. Note: a Tukey HSD post hoc did not find any significant differences between groups ($\alpha = 0.05$).

The normalized lumbar flexion values calculated in this investigation revealed a three-way interaction between gender, mass and time ($p = 0.0451$). The normalized lumbar flexion values recorded for both gender groups and the light, average and heavy body mass groups were independently ran through a one-way GLM with time as a factor. Only the heavy body mass group showed a time main effect ($p = 0.0308$). The time-varying response of the heavy body mass group displayed a significant decrease in normalized lumbar flexion after 120 minutes of driving (58.6 (15.3) degrees) versus the first the 30 minutes of driving (55.9 (15.1) degrees) (Figure 4.5a).

Males and females within the light, moderate, and heavy body mass groups were then individually though a one-way GLM with time as a factor. Of the six individual groups, only males in the heavy body mass group displayed a significant time effect (p =

0.0193). The time varying response of the males, in the heavy body mass group showed significantly less (≈ 5º) normalized lumbar flexion after 120 minutes of driving (66.2 (15.8) degrees) versus the first the 30 minutes of driving (61.1 (17.1) degrees) (Figure 4.5b).

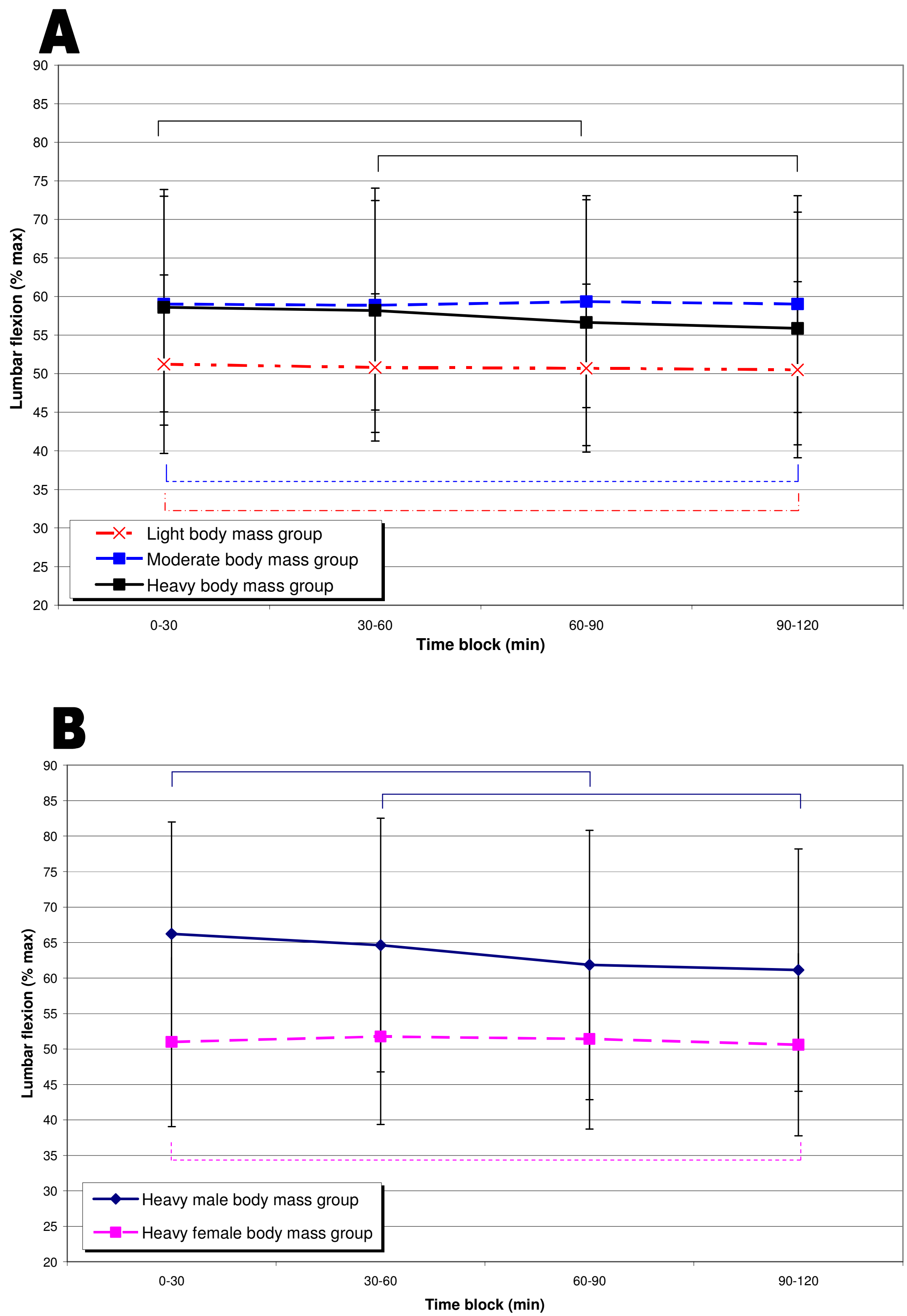



**Figure 4.5:** Time varying responses of normalized lumbar flexion values for the light, moderate, heavy (A), heavy male and heavy female body mass groups (B). Each variable was averaged across all participants for each 30 minute time interval. Values grouped under the same horizontal bars are not significantly different from each other ($\alpha = 0.05$).

#### 4.3.3 *Joint and segment angles*

No statistical differences or interactions between body mass and gender groups was observed for all joint and segment angles. Statistical differences with respect to time were seen for both hip angle ($p = 0.0358$) and trunk angle ($p = 0.185$). Hip angle statistically increased from 106.7 (7.9) to 108.4 (7.9) degrees after 45 minutes. Trunk angle increased from 114.5 (7.7) to 115.5 (7.5) degrees after 30 minutes. No interactions between gender or body mass with time were observed.

#### 4.3.4 *Gender and body mass response: CoM, Hip and CoP locations during driving*

No significant differences with respect to time, gender, or mass was observed in the HAT CoM, hip, and CoP locations relative to the front of the seat pan. The location of the HAT CoM with respect to the hip, the hip with respect to the CoP, and the CoP with respect to the front of the seat pan, were located posterior to each other, respectively across all body mass categories (Figure 4.6).

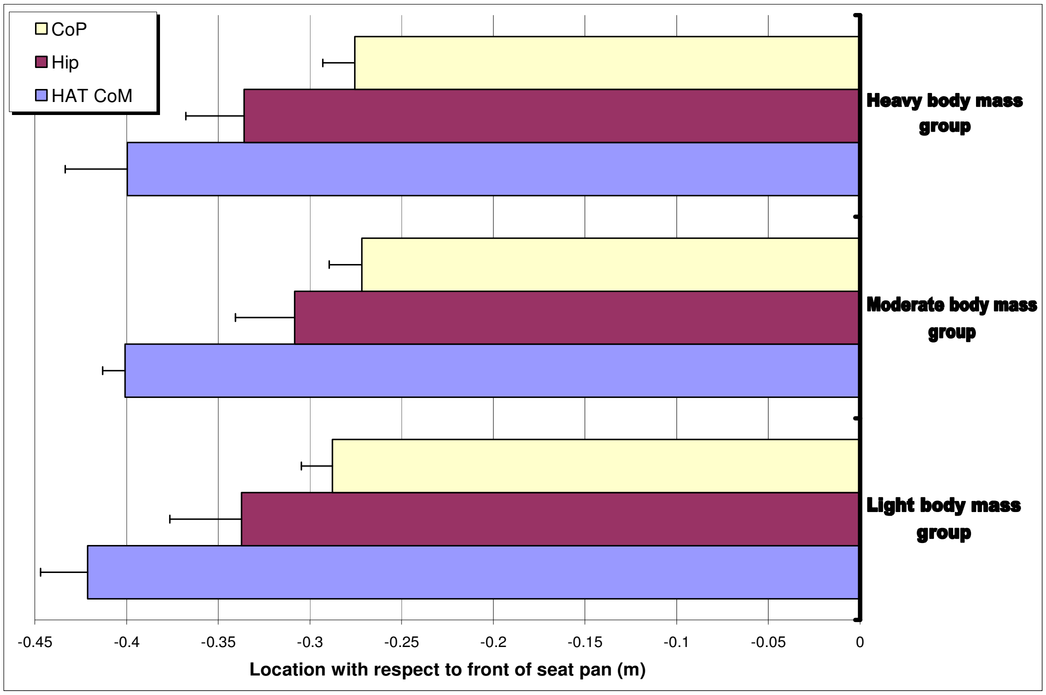



**Figure 4.6:** Mean location of HAT CoM, hip, and CoP with respect to the front of the seat pan over a 120 min driving simulation, for the light, moderate and heavy body mass groups (n = 15).

### 4.4.0 Self reported discomfort measures

No statistically significant gender or body mass main effects were found for the self reported ratings of discomfort. There were also no interactions to report. Of the 20 body regions recorded over the 120 minute driving simulation, 8 body regions showed a significant increase in discomfort over time. All body regions displayed some increase in discomfort over time. The 8 body regions displaying a significant increases in discomfort over time included the neck ($p = 0.0019$), left shoulder ($p = 0.0045$), right shoulder ($p = 0.0075$), left upper back ($p = 0.0228$), right upper back ($p = 0.0173$), middle back ($p = 0.0316$), the lower back ($p = 0.0028$), the left buttock ($p = 0.0005$) and the right buttock ($p = 0.0017$). Overall body discomfort also increased significantly with respect to time ($p = 0.0027$). Of the 8 body regions showing significant time varying response, 7 displayed a significant change in discomfort after 60 minutes (Table 4.3).

**Table 4.3:** Means and standard deviations of a driver's self reported ratings of perceived discomfort. Values are expressed in millimeters, with 0 mm corresponding to "no discomfort" and 100 mm corresponding "extreme discomfort".

| Body Region | (n = 24) | 0 min | 30 min | 60 min | 90 min | 120 min |
|---|---|---|---|---|---|---|
| Neck$^{ST}$ | Mean | $0.3^{a}$ | $3.3^{a,b}$ | $3.7^{a,b}$ | $5.5^{b}$ | $6.1^{b}$ |
| | *SD* | *1.1* | *4.3* | *4.8* | *5.9* | *9.0* |
| (L) Shoulder$^{ST}$ | | $0.0^{a}$ | $1.8^{a,b}$ | $4.1^{b}$ | $4.6^{b}$ | $5.4^{b}$ |
| | | *0.0* | *4.8* | *6.4* | *7.4* | *8.4* |
| (R) Shoulder$^{ST}$ | | $0.3^{a}$ | $2.4^{a}$ | $4.0^{a,b}$ | $7.5^{b}$ | $7.8^{b}$ |
| | | *0.7* | *6.2* | *8.0* | *10.9* | *13.8* |
| (L)Upper Back | | $0.5^{a}$ | $1.0^{a}$ | $1.5^{a}$ | $2.0^{a,b}$ | $4.8^{b}$ |
| | | *1.9* | *2.6* | *3.6* | *3.6* | *7.7* |
| (R)Upper Back$^{ST}$ | | $0.0^{a}$ | $1.9^{a}$ | $3.0^{a,b}$ | $4.6^{a,b}$ | $7.4^{b}$ |
| | | *0.0* | *6.3* | *7.9* | *10.4* | *13.4* |
| Middle Back$^{ST}$ | | $1.0^{a}$ | $2.2^{a}$ | $4.0^{a,b}$ | $5.0^{a,b}$ | $8.9^{b}$ |
| | | *2.3* | *4.5* | *10.1* | *8.1* | *14.4* |
| Lower Back$^{ST}$ | | $1.4^{a}$ | $3.4^{a}$ | $7.1^{a,b}$ | $10.1^{b}$ | $10.1^{b}$ |
| | | *2.9* | *5.0* | *11.4* | *10.7* | *11.6* |
| (L) Side of Body | | 0.3 | 0.9 | 2.1 | 2.2 | 3.6 |
| | | *1.2* | *3.5* | *9.8* | *5.2* | *6.6* |
| (R) Side of Body | | 0.0 | 1.3 | 2.0 | 4.3 | 4.8 |
| | | *0.0* | *6.5* | *8.8* | *11.6* | *12.8* |
| (L) Upper Pelvis | | 0.3 | 0.5 | 0.5 | 0.8 | 1.7 |
| | | *1.1* | *1.8* | *1.3* | *2.2* | *4.3* |
| Sacrum/tail bone | | 0.7 | 0.5 | 0.9 | 1.0 | 2.2 |
| | | *2.2* | *1.7* | *2.6* | *3.0* | *5.4* |
| (R) Upper Pelvis | | 0.3 | 0.5 | 1.2 | 1.0 | 1.7 |
| | | *1.3* | *1.6* | *2.9* | *3.2* | *4.7* |
| (L) Buttocks$^{ST}$ | | $0.5^{a}$ | $0.6^{a}$ | $1.0^{a}$ | $2.5^{a}$ | $8.2^{b}$ |
| | | *1.9* | *2.1* | *2.7* | *5.3* | *9.3* |
| (R) Buttocks$^{ST}$ | | $0.1^{a}$ | $0.2^{a}$ | $0.6^{a}$ | $2.7^{a}$ | $7.5^{b}$ |
| | | *0.4* | *0.6* | *1.7* | *5.7* | *9.7* |
| (L) Upper Thigh | | 0.0 | 0.3 | 1.2 | 2.1 | 4.9 |
| | | *0.0* | *1.4* | *3.5* | *6.3* | *10.9* |
| (R) Upper Thigh | | 0.4 | 0.7 | 1.5 | 1.5 | 3.9 |
| | | *1.8* | *2.3* | *4.1* | *5.3* | *9.5* |
| (L) Lower Thigh | | 0.0 | 0.2 | 0.9 | 2.0 | 3.7 |
| | | *0.0* | *0.8* | *2.9* | *5.8* | *11.4* |
| (R) Lower Thigh | | 0.0 | 0.3 | 1.1 | 1.0 | 2.6 |
| | | *0.0* | *1.4* | *3.7* | *4.0* | *9.3* |
| (L) Side of Leg | | 0.0 | 0.5 | 0.5 | 2.1 | 3.4 |
| | | *0.0* | *1.8* | *1.4* | *5.9* | *10.9* |
| (R) Side of Leg | | 0.0 | 0.4 | 0.7 | 1.8 | 2.3 |
| | | *0.0* | *2.0* | *2.5* | *5.7* | *7.3* |
| Summed Body Discomfort$^{ST}$ | | $5.9^{a}$ | $22.7^{a,b}$ | $41.7^{a,b}$ | $64.4^{b,c}$ | $101.0^{c}$ |
| Score out of 2000 | | *9.0* | *23.0* | *45.0* | *73.2* | *124.4* |

$^{ST}$ indicates variables changed significantly over time ($p < 0.05$). Values labeled with different letters e.g. $^{a,b,c\ldots}$ are significantly different from one another based on the results of a Tukey HSD post hoc ($\alpha = 0.05$).

## 5.0 Discussion

In most of the current literature, stature and gender have been the primary independent variables tested in a driving situation, with the influence of body mass generally left untested. Stature and gender were both controlled in this investigation, which gives a clearer picture of how body posture, pressure distribution and discomfort are influenced by body mass during a prolonged driving situation. This information provided insight to influence of body mass as a risk factor in the development of driver discomfort in prolonged driving situation.

### 5.1.0 Interface pressure distribution recordings

#### *5.1.1 Body mass effects*

Agreeing with the hypothesis in this investigation, heavier participants were shown to produce increased total ischial tuberosity (IT) and total seat pan/back pressures during driving. However, when normalized to seat contact area, differences in seat pan IT pressures and total seat back pressure profiles disappeared ($\alpha = 0.05$) across body mass groups. Mergl, et al (2005) has shown that a seat contact pressure of 20 kpa ($\approx$ 140 mmHg/cm$^2$) in the buttocks region is the predictive threshold of buttock discomfort during driving. The mean normalized IT pressure readings and total seat back pressures in this investigation both remained below 140 mmHg/cm$^2$, while the self reported assessments of perceived discomfort in the buttock region were less than 10 out of 100 mm, which is considered minimal discomfort during driving (Mergl et al., 2005). The ability of larger body mass groups to disperse pressure over greater areas indicates that drivers, regardless of body mass, are able to preserve capillary blood flow in regions of high pressure (Landis, 1931; Le et al., 1984; Peters and Swain, 1997; Swain, 2005). This limits the influence of body mass on the development of discomfort during driving (Kolich et al., 2000; Landis, 1931; Le et al., 1984; Mergl et al., 2005). These results show that regardless of body mass, with healthy tissues and a deformable seat pan, the elevated seat pan total IT pressures and seat back total pressures produced by heavier participants are dispersed over greater areas, limiting the development of discomfort in a prolonged driving situation.

Seat pan total pressure profiles across body mass groups did not produce the same trends as the seat back and IT pressure profiles during driving. The time varying response of seat pan total pressure recordings for the heavy and moderate body mass groups were similar to each other, but different than the time varying response of the light body mass group. The light body mass group displayed a unique time varying response relative to the moderate and heavy body mass groups. All groups responded in a similar fashion for the first 60 minutes. Following 60 minutes, the light body mass group displayed a decreased rate of change in seat pan total pressure relative to heavier participants. Researchers from the Woodbridge Foam Corporation® have shown that increasing the applied load to a material property such as seat foam will induce an increase in seat deformation (Wilson and Blair, 1994). The stiffness response or damping of the material properties of car seat foam were also shown to be proportional to the applied load (Wilson and Blair, 1994). In a damped system, the response of the light body mass group would reach a state of deformation equilibrium sooner than the heavier participants, explaining why the total pan pressures showed a decreased response after 60 minutes relative to heavier participants.

The time varying response of seat pan total PPA showed that the heavy body mass groups displayed elevated seat pan total PPA relative to the moderate and light body mass groups, at every time interval, over the entire driving simulation. Also, the time varying response of the seat pan total PPA change showed an increased rate of change for the heavy body mass group relative to the moderate and light body mass groups.

The literature indicates that the heavy body mass group would have possessed characteristically larger and wider hips relative to the moderate and light body mass groups (Byers, 2002). Looking at a 1 frame recording of a heavy participant's pressure profile; the interface pressure recordings for the seat pan show that heavier populations are not adequately accommodated with the design of a typical seat pan (Figure 5.1). With the heavy body mass group covering the surface area of the seat pan beyond its lateral boundaries, the ability of this population to distribute pressure though deformation of body tissues would be restricted relative to the moderate and light body mass groups. Previous literature has reported that a typical car seat is designed to accommodate populations in the $50^{th}$ percentile (Kolich and Taboun, 2002). The literature has also shown that as populations deviates from the anthropometrics of a $50^{th}$ percentile population, proportional increases in discomfort onset and magnitude follow (Gyi and Porter, 1999; Porter and Gyi, 1998). By the heavy body mass group not being accommodated by the seat pan, heavy body mass groups are at greater risk of developing musculoskeletal disorder during prolonged driving (Kolich and Taboun, 2004; Porter and Gyi, 1998; Porter and Gyi, 2002; Thomas et al., 1991).

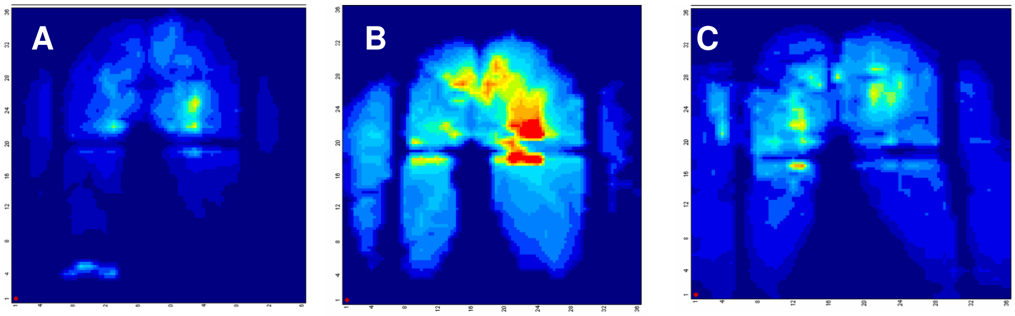



**Figure 5.1:** Seat pan pressure profiles of a typical light (A), moderate (B) and heavy (C) body mass group during prolonged driving. Note that the pressure profile of the heavy body mass group is cut at the lateral boarders of the seat pan.

All body mass groups did not produce statistically different ($\alpha = 0.05$) total IT PPA, but did while produce differences in total set pan PPA values. This may partially explain why the heavy body mass group did not produce significantly elevated ratings of perceived discomfort. The dispersion of peak IT pressures over an area may influence discomfort to a greater extent than total seat pan pressure distribution. Although the ratings of perceived discomfort showed no differences between groups, decreasing a driver's ability to disperse pressure over area likely elevates their probability of developing an ischemic state (Swain, 2005), increasing their risk of developing discomfort during a prolonged driving situation (Kolich et al., 2000; Porter et al., 2003).

### *5.1.2 Gender effects*

Females in this investigation displayed lower right IT pressures per unit area (PPA) relative to males, while producing equivalent left IT PPA. Looking closer to the IT PPA trends observed in this investigation (Appendix E and F), it is apparent that these statistical findings are likely influenced by the response of the light and moderately massed female body mass groups. The light and moderate female body mass groups displayed lower right PPA recordings in the first 15 minutes of driving relative to the heavy female body mass group and male group, with a more pronounced difference in the last 15 minuets of driving. It would be assumed that the light and moderately massed female body mass groups would produce increased left IT PPA relative to males and the heavy female body mass group, to account for their decreased right IT PPA. This trend however was not observed. In the first 15 minutes of driving, the heavy female body mass group displayed a trend of higher left IT PPA relative to males (Appendix E and F), while producing similar right IT PPA, these trends however were not consistent across the 120 minute driving simulation, as the difference in left IT PPA regressed to the mean of the male groups left IT PPA recordings over the 120 minute driving simulation. It is likely that the small sample sizes of the gender specific body mass groups (n=4) prevented these observed trends in the female body mass groups from attaining statistical significance.

Females have been shown to posses increased soft tissue in the buttock region (Zarcharkow, 1988) and wider ischial tuberosities (Byers, 2002; Krogman, 1962). Differences in anthropometrics between males and females may give these populations the ability to disperse pressure over a seating surface differently (Swain, 2005). This may explain why light and moderately massed females displayed lower right IT pressure per unit area values as defined by this investigation (all of the cells adjacent to and including the IT pressure value within 10% of the defined peak IT pressure cell) relative to the male and heavy female population during driving. An increased total seat pan area, in association with a decreased right IT total pressure suggests that light and moderately massed females disperse their right IT pressures over greater areas, keeping the cells adjacent to the defined peak pressure cell below the 10% IT pressure threshold. Understanding why the heavy female body mass group did not display the same trends as the lighter female body mass groups may be attributed to the exaggerated pelvic geometries characteristic of a heavier population (Byers, et al 2002). It has also been shown by forensic scientists that as a female's mass and girth increases, the sex specific characteristics of the female skeleton begin to resemble that of a male skeleton (Byers, et al 2002). This could mean that as females become larger, their pelvic geometries begin to resemble that of a male population. Heavy females may then disperse IT pressure over a seat pan in a similar fashion as male populations. This may explain why the IT PPA response of the heavy female group was more characteristic of the male population versus the lighter female body mass groups used in this investigation.

These findings are partially supported by the literature, where males have been shown to produce decreased mean right total IT pressures relative to females while driving (Coke et al., 2007; Gyi and Porter, 1999). It should also be noted that the body masses of the female populations used in previous investigations ($5^{th}$ (Gyi and Porter, 1999) and $50^{th}$ percentile (Coke et al, 2007) female populations) were similar to the body masses of the moderate and light body mass groups.

With the results from literature and results from this investigation showing that females tend to produce an relatively lower IT pressures on the right IT's relative to their left IT's during driving suggests that gender specific interactions with the brake and accelerator pedal may have also influenced their IT pressures. Without measurements of brake and accelerator force, seat/tissue deformation during driving, or external pelvic measures, the precise role body composition, anthropometrics, and/or gender specific interactions with the accelerator and break pedals contributed to male and female differences in right IT PPA distributions during driving is left relatively unknown. It can be stated however that males and moderately massed and lightly massed females distribute IT pressures differently during driving, even when anthropometric variables like stature and body mass are controlled between gender groups. No clear trends across body mass groups and genders in the ratings of perceived discomfort (Appendix C and D). Therefore, clear associations between discomfort and the affect an asymmetric IT PPA pressure distribution cannot be made. Lack of significance or clear discomfort trends between the male and female populations may be due to the complex nature of discomfort perception (de Looze et al., 2003). Peak pressure are not be the sole contributor to discomfort while driving, though it has been shown to significantly influence discomfort (Gyi and Porter, 1999; Kolich et al., 2000; Kolich, 2003; Mergl et al., 2005; Porter et al., 2003).

**5.2.0 <u>Kinematics</u>**

Males, females and all body mass groups displayed statistically equivalent CoM, and CoP locations over the 120 minute driving simulation. These findings agree with the literature, showing that when assuming a driving posture, all people regardless of sex or body mass adopt a posteriorly rotated pelvic posture during prolonged driving (Beach et al 2007;Coke et al., 2007, Reed et al 1991). This was also verified by the obtuse hip (≈108º) and trunk angles (≈115º) observed by participants across body mass groups (Table 5.1). Looking to previous literature, males and females have been shown to position their HAT CoM differently during driving (n = 24) (Coke et al., 2007). It is acknowledged that the decreased degrees of freedom used in this investigation (n = 15 vs n = 24) may have prevented differences in CoM and CoP position from being observed between genders and/or body mass groups. It should be noted however that the investigation reporting differences in HAT CoM body positioning while driving contained stature differences between gender groups (Coke et al., 2007). Previous investigators have shown that a person's posture during driving is greatly influenced by the geometry of the seat (Coke et al., 2007), and the suitability of a car's interior geometry to accommodate a driver's stature (Porter et al., 2003). Controlling for both stature and seat adjustability would likely explain why no statistical differences in body position was observed across body mass and gender groups.

All joint and segment angles calculated in this investigation were similar to the ranges recorded in the literature (Coke et al., 2007; Grandjean, 1980; Gyi et al., 1998; Park et al., 2000; Rebiffe, 1969) (Table 5.1).

**Table 5.1:** Comparisons of observed joint and segment angles (degrees) and the published values as reported in the literature. All data is expressed as the mean of all participants, across all time intervals, for the 120 minute driving simulation.

| **Classification** | **Rebiffe (1969)** | **Grandjean (1980)** | **Porter & Gyi (1998)** | **Park (2000)** | **Coke (2006)** | **Observed *(n = 24) Mean (SD) Range*** |
|---|---|---|---|---|---|---|
| Hip Angle | 95-120 | 100-120 | 90-115 | 103-131 | 103 - 118 | *108.1 (7.9)*<br>*91 - 119* |
| Knee Angle | 95-135 | 110-130 | 99-138 | 120-152 | 121- 134 | *121.6 (6.6)*<br>*106 - 135* |
| Trunk Angle | N/A | N/A | N/A | N/A | 104 - 118 | *115.4 (7.5)*<br>*94 - 125* |

There were no statistical differences with respect to joint and segment angles for either gender or body mass group in this investigation. These results show that when seat back angle and stature are controlled, gender and body mass has little influence on the two-dimensional joint and segment angles adopted by participants during driving. These results are supported by previous investigations controlling for stature (Reed et al., 2000), and by statements of other researchers in the car seat literature (Chaffin et al., 2000).

The observed time varying responses in hip and trunk angles were approximately 1°

over 120 minutes of driving. With all body mass and gender groups displaying similar increases over time, it is likely that variables such as segment rotations, changes in torso posture, deformation of the car seat's foam properties and active marker movement contributed to the observed increases in hip and trunk angle during prolonged driving.

### *5.2.1 <u>Pelvic kinematics</u>*

The recorded pelvic angles in this investigation are comparable to results previously published in the automotive literature (Beach et al, 2007; Coke et al, 2007). From this investigation it was observed that females rotated their pelvis with respect to the vertical to a lesser degree than males when adopting a posture for driving. These findings are supported by previous trends reported in the automotive literature (Beach et al, 2007). It is thought that by controlling for both mass and stature, decreased inter-participant variability would be attained. The use of tri-axial instead of uni-axial strain gage accelerometers to measure pelvic angle provides the ability to measure cross-talk between acceleration axes, which would increase measurement resolution. Increased resolution and decreased inter-participant variability may have allowed statistical differences in pelvic angle with respect to vertical between males and females to be observed during this investigation.

The increased posterior pelvic rotation displayed by males during driving may be associated with a decrease in hamstring flexibility (Beach et al 2007, Youdas et al, 2005). It is thought that because males posses tighter hamstrings, they will increase the rotation of their pelvis in the posterior direction to decrease the strain or stretch imposed on their hamstring complex when in a sitting posture such as driving (Beach et al, 2007).

Postural differences with respect to pelvic rotation combined with an array of other variables such as lumbar posture and tissue loading may have convoluted the perception of lower back discomfort and pelvic discomfort (de Looze et al., 2003; Helander and Zhang, 1997), preventing gender specific ratings of perceived discomfort from being recorded (Appendix B). With males and females displaying unique biomechanical responses in pelvic orientation when placed in driving postures, these populations will be exposed to different sacroiliac and lumbar loading patterns (Mitchell and Mitchell, 1999; Vleeming, 1997). Gender is then considered as a risk factor in the development of discomfort during prolonged driving.

Results have shown that regardless of body mass, the pelvis will be placed into posterior rotation relative to standing while driving. These results support previous findings in the car seat literature that have recorded pelvic rotation from standing to a driving posture (Beach et al, 2007, Coke et al., 2007; Reed et al., 1991). The moderate body mass group in this investigation displayed an increased posterior pelvic rotation with respect to the vertical axis relative to the light and heavy body mass groups during driving. These results suggest that moderately massed participants display unique pelvic postures relative to other body mass populations. It was hypothesized that an increase in body mass would produce an increase in posterior pelvic rotation during driving. It was thought that with an increase in HAT mass, larger moments would then be placed on the lumbar spine during driving, encouraging the pelvis to posteriorly rotate, as the spine would be placed into greater degrees of lumbar flexion (Mitchell and Mitchell, 1999; Vleeming, 1997).

Though these results showed that the moderate body mass group displayed statistically different pelvic angles relative to the other body mass groups, a Tukey post hoc analysis ($\alpha = 0.05$) revealed that these populations were not statistically different from each other. A Tukey post hoc is a conservative statistical comparison tool, less likely to make type 2 errors (Kuehl, 2000). The increased variability of body composition, pelvic geometry and standing postures could have all contributed to this non-significant Tukey post hoc. Currently no relevant literature has stated that moderately massed participants display unique pelvic geometries, body composition characteristics or lumbo-sacral kinematics from sit to stand relative to other body mass groups. Also, there were no other transducers used in this investigation recording differences in pelvic postures across body mass groups during driving. Confident conclusions stating that the pelvic angles of a moderately massed body mass group are different from the pelvic angles of a light or heavy body mass group are tentative.

### *5.2.2 **Lumbar kinematics***

During driving, regardless of body mass, all groups displayed increased normalized lumbar flexion when moving from a standing to a driving posture. Similar changes have been documented in the literature where the normalized lumbar angles attained during driving were between 40% and 60% of maximum flexion (Beach et al, 2007, Coke et al., 2007; Reed et al., 1991). These angles all lie within the initiation range (≈35-40% max flexion) of flexion-relation (FR) during sitting (Callaghan and Dunk, 2002; O'Sullivan et al., 2006). If these spinal angles induce FR across seating environments, the normalized lumbar flexion angles recorded in this investigation would lead to the conclusion that the moments created by a driver's HAT segment are supported by the posterior passive tissues of the lumbar spine (Callaghan and Dunk, 2002; O'Sullivan et al., 2006). These fixed flexed postures would also cause the passive tissues of lumbar spine to creep (McGill and Brown, 1992), theoretically placing the spine into further flexion as time progressed.

The time varying response of males in the heavy body mass group were the only group of the six gender/mass groups that displayed significant changes in normalized lumbar flexion. The heavy male body mass group was shown to decrease the flexion of their lumbar spine (≈ 5°) as driving time progressed. These results are not supported by the car seat literature (Beach et al, 2007, Coke et al., 2007; Reed et al., 1991), and conflicted with the hypothesis that the lumbar the spine would be placed into greater degrees of spinal flexion as creep progressed in the passive tissues of the lower back during driving (McGill and Brown, 1992).

The straightening response displayed by the heavy male population in this investigation occurred after 60 minutes of driving. It is possible that the 60 minute collection interval used by Coke, et al (2007) and the 10 minute collection interval used by Beach, et al (2007) may not have been long enough to observe these trends. Also, the use of a sonic digitizing probe by Reed, et al (1991) likely did not possess the sensitivity needed to record the small changes in lumbar flexions displayed by heavy male population in this investigation.

Looking to reports in the office seating literature, a similar straightening response of a male population was observed after 1 hour of sitting; males were shown to adopt a less flexed lumbar posture in the second hour of sitting while females maintained the same relative spinal angles (Beach et al., 2005). Comparing between investigations, the mean mass of the male population from Beach, et al (2005) (76.8 (15) kg) was similar to the heavy male population used in this investigation (83.7 (0.73) kg). This suggests that Beach, et al (2005) may have observed a heavy male response of decreased lumbar flexion over time and not solely a gender response as reported. With no control of the body mass used in Beach, et al (2005) investigation, differences in seating environments and the use of different kinematic methods to record lumbar posture, this statement can only be left to speculation.

As stated by Beach, et al (2005) the proposed mechanism for this reduction in normalized lumbar flexion over time may have been related to associated increases in spinal stiffness recorded in their investigation. It was proposed that undetected gender specific lumbar flexion/extension movements during sitting may have increased fluid absorption in the intervetebral discs of men, resulting in increased intervetebral disk height, increasing stiffness and decreased lumbar flexion after 1 hour of sitting (Beach et al., 2005). It is possible that increases in intervetebral disk height may have caused the straightening response of heavy males in this investigation, but measurements of lumbar stiffness need to be conducted before these findings can be confirmed in a driving situation.

An alternate possible mechanism explaining the decreased lumbar flexion response of the heavy male population in this investigation may be attributed to the inherit flexibility differences reported between males and females in the literature (Beach et al 2007, Youdas et al, 2000). It has been reported in the literature that males posses a decreased flexibility in the lumbo-pelvic region (Beach et al, 2007), and the hamstring complex (Youndas et al, 2000). When looking at the time varying trends of the pelvis with respect to vertical for the heavy male and heavy female body mass groups; the heavy male group showed trends of decreased pelvic rotation over time, while the heavy female body mass group did not display the same trend (Figure 5.2). The time varying anterior rotation of the pelvis during driving illustrates that hamstring flexibility does not entirely define the posture of the pelvis during driving. Documented decreases in flexibility observed in regions other than the hamstring complex, like the lumbo-sacral region (Beach et al, 2007) likely influence the orientation of the pelvis as driving time progressed.

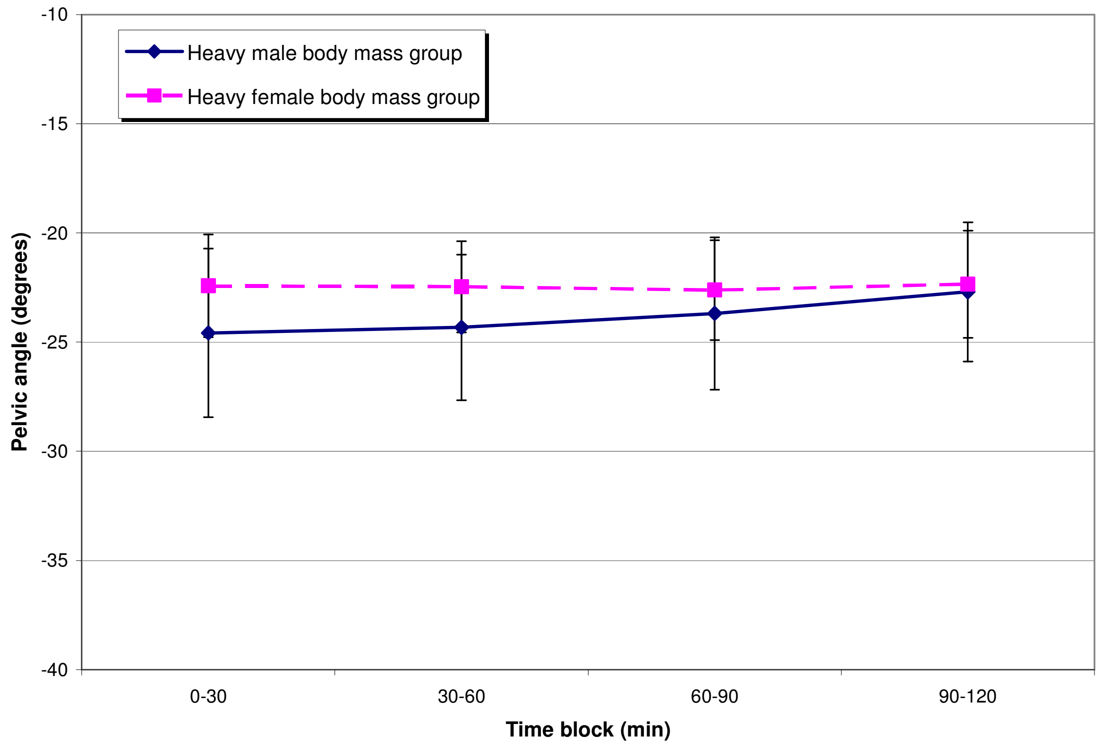



**Figure 5.2:** Time varying trend of pelvic angle with respect to vertical of the heavy male and heavy female body mass groups. Each variable was averaged across participants, for each 30 minute time interval, over the entire 120 minute driving simulation.

The increased posterior pelvic rotation trend displayed by the heavy male population would have likely been initially influenced by a decrease in hamstring flexibility characteristic of a male population (Youndas et al, 2005). The initial posterior orientation of the male pelvis to accommodate a decreased flexibility in the hamstring complex would place added strain or stretch on the posterior muscles of the lumbar region, which are also characteristically less flexible in a male population (Beach et al, 2007). The stretch reflex in the posterior muscles such as the erectors or multifidus muscles may then have been triggered when stretched past their resting lengths (Nicol et al, 2006), inducing the straightening response observed in the lumbar spine and pelvis by the heavy male population.

Explaining why this response only occurred in the heavy male population after 1 hour of driving may be attributed to larger moments a heavier HAT segment would place on the posterior muscles of the lumbar spine. The increased applied moment placed upon the characteristically tighter male lower back musculature (Beach et al 2007) may have increased the low level static muscle activations experience by the posterior muscles of the lower back during prolonged driving (Grieco, 1986; McGill, 2005). Increasing the muscle activation of the lower back muscles during driving may then induce fatigue over time in the lumbar musculature (Jorgensen, 1988). The stretch reflex in skeletal muscle has been shown to be elevated as fatigue is induced (Nicol et al, 2006), which may explain the lumbar and pelvic straightening response observed after 1 hour of driving in the tight lumbo-pelvic region of a heavy male population.

It should also be noted that when documenting a decreased lumbo-pelvic flexibility in a male population, the male population used was 84.2 (4.5) kg (Beach et al, 2007), which was similar to the mass of the heavy male population used in this investigation (83.7 (0.73) kg). The documented decreases in lumbo-pelvic flexibility observed by the male population in Beach et al, (2007) may then been a heavy male response. Without recordings of male lumbo-pelvic flexibility across different body mass groups, these statements can only be left to speculation. Electromyographical recordings of the lumbar musculature and hamstring complex are required during a driving situation before this proposed mechanism can be confirmed.

## 6.0 Conclusions

This study showed that in a typical car seat, heavy body mass groups do not have the same ability to disperse pressure over the seat pan as a moderate and light body mass group. In a typical car seat, increased body mass increases a driver's risk of developing discomfort in a prolonged driving situation.

The time varying response of the heavy male body mass group's normalized lumbar flexion was shown to display a decrease in normalized flexion over time. The response may be due to an increase in lumbar stiffness or a stretch-reflex response of the muscles in the lumbar region. Without appropriate measurements to verify this hypothesis, future investigations are needed to verify these proposed mechanisms.

Overall, the results from this investigation show that in a typical car seat, participants of increased body mass may be exposed to increased seat pressures. Heavy populations are then exposed to an increased risk of discomfort during prolonged driving. Appropriate design of the seat pan to accommodate the increased applied pressure areas produced by heavier participants would reduce their risk of discomfort during prolonged driving.

When controlling for stature, seat position and body mass, no statistical differences in lower body joint/segment angles and body positioning were observed. These results show that previous investigations reporting gender differences in joint/segment and body position during driving are likely associated to differences in stature between gender groups or to gender specific seat adjustments that influence their driving postures.

Biomechanical differences in IT pressure, lumbar posture and pelvic posture all show that males and females are exposed to different forces during driving. Males and female populations may then progress though different discomfort pathways during driving. With control of anthropometric variables such as body mass and stature, confident conclusions that gender is a risk factor in the development of discomfort during prolonged driving can be made.

Overall, these results have shown that even when controlling for population characteristics such as body mass and stature, no gender specific differences in two-dimensional kinematic joint, segment and body posture were observed during driving. However, biomechanical differences in both pressure distribution and lumbo-sacral postures provide substantial evidence that gender is a risk factor in the development of discomfort during prolonged driving.

## 7.0 Future directions

Measurements of lumbar stiffness and lumbar and hamstring muscle activation must be made if the loading characteristics of the lumbar region can be better understood in driving situations. Development of a method capable of recording flexion/extension movements of the lumbar spine may provide better insight to possible gender differences in intervetebral nutrition during driving. This information may help understand the possible mechanisms attributed to the straightening response observed by heavy males after 1 hour of prolonged driving.

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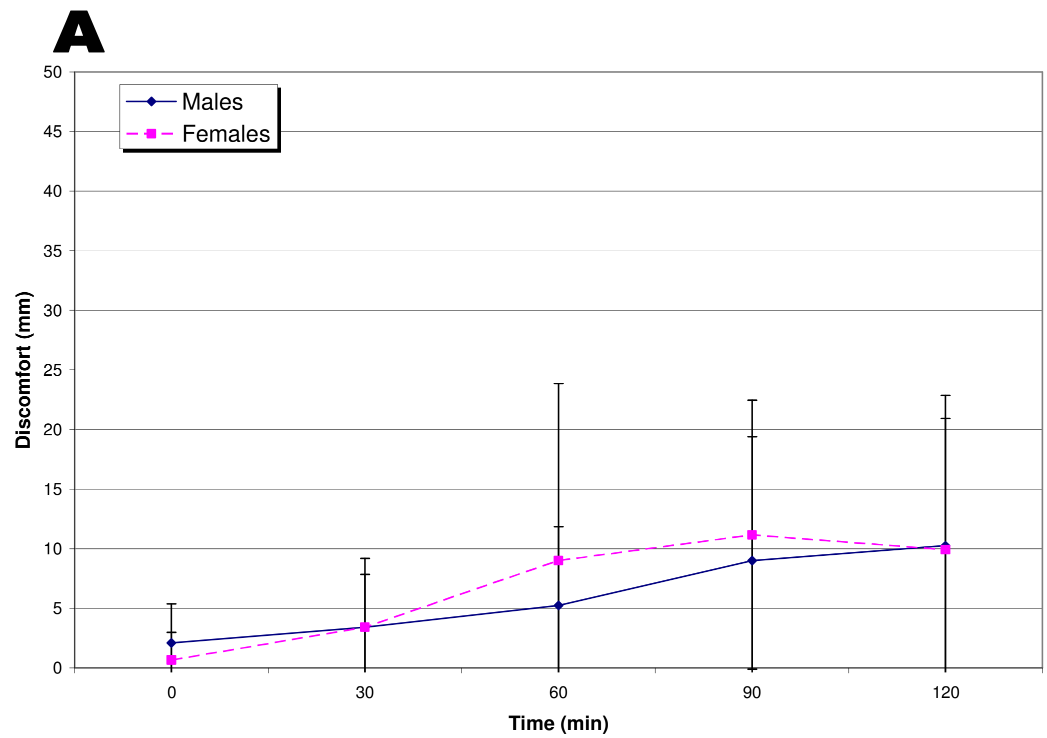
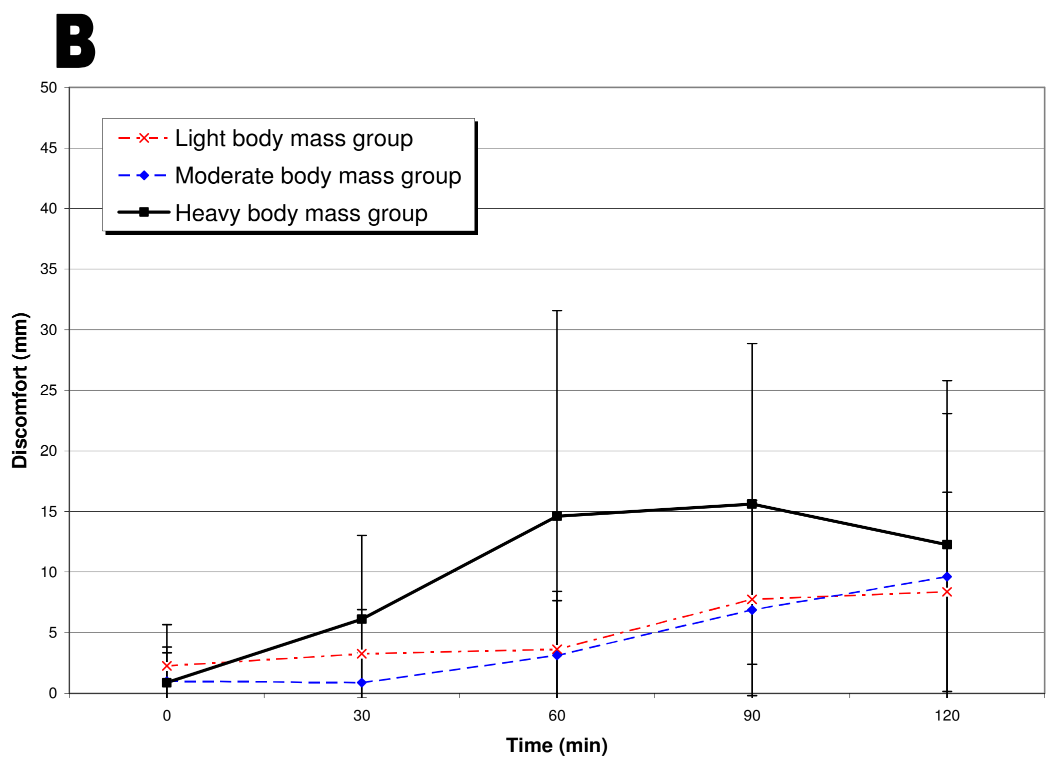
**Appendix A:** Method of finding the anatomical landmarks for placement of kinematic markers (Magee, 1992).

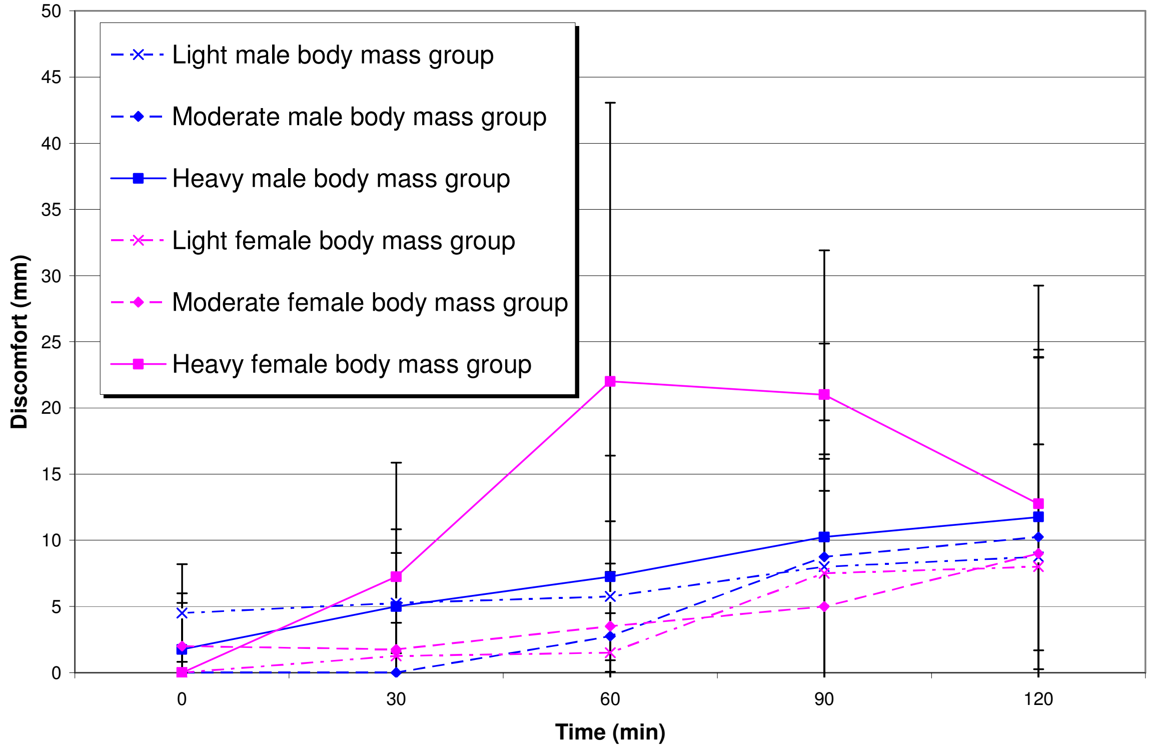
| | |
|---|---|
| **Ulnar Styloid** | In anatomical position, the Ulnar styloid is the bony protrusion located on the medial/posterior side of the ulna. |
| **Lateral Epicondyle (Humerus)** | Palpate the lateral side of the distal end of the humerus (2 "bumps" on the distal end). |
| **Acromium** | Find the clavicle (located just anterior to the sternum) and follow it laterally until a "bump" is felt at its distal end. |
| **Targus** | Skin covering the ear canal |
| **Greater Trochanter** | Boney landmark distal to the hip, along the line of the iliotibial band, separating the quadriceps and gluteus complexes. |
| **Lateral Epicondyle (Femur)** | Palpate the lateral side of the distal end of the femur, located at the origin of the lateral collateral ligament. |
| **Lateral Malleolus (Ankle)** | Palpate the lateral side of the distal end of the fibula (Large protuberance on the lateral side of the ankle). |

**Appendix B**: Lower back region discomfort trends. Trends are broken into male and female populations (A), light, moderate and heavy body mass groups (B), and light male/female, moderate male/female, and heavy male/female body mass groups (C).



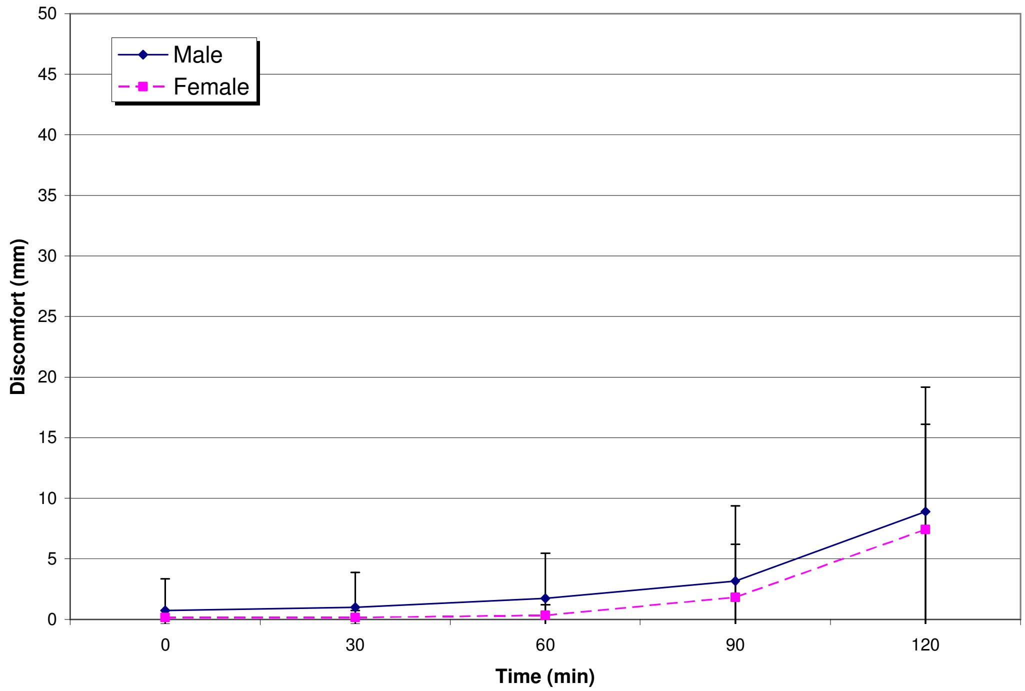



c

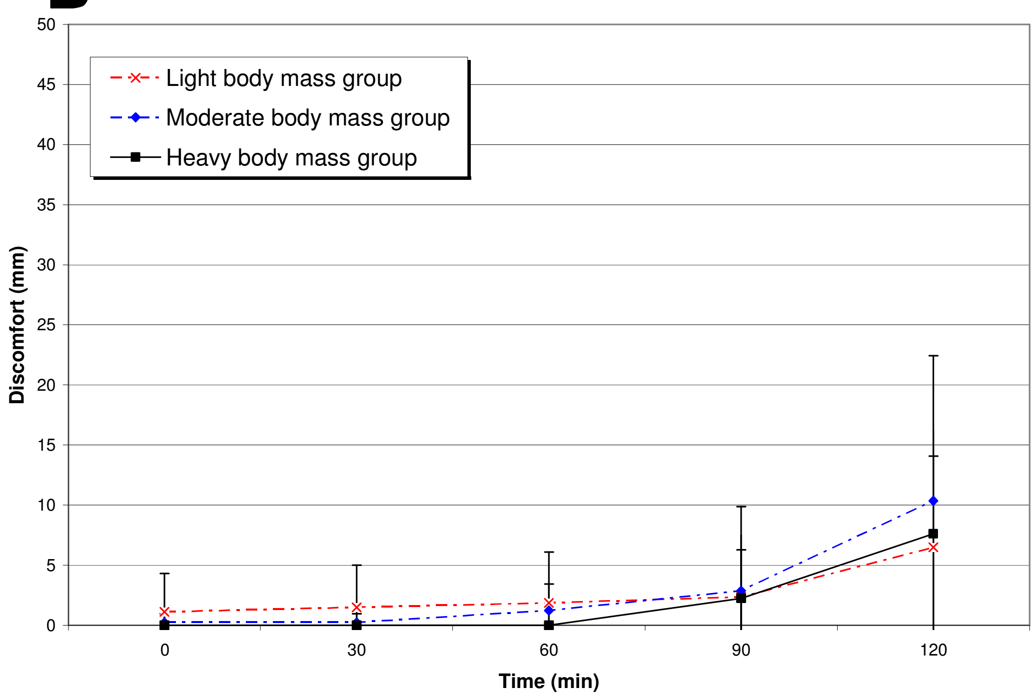

50
45
40
35
30
25
20
15
10
5
0
Discomfort (mm)
Light male body mass group
Moderate male body mass group
Heavy male body mass group
Light female body mass group
Moderate female body mass group
Heavy female body mass group
0
30
60
90
120
Time (min)

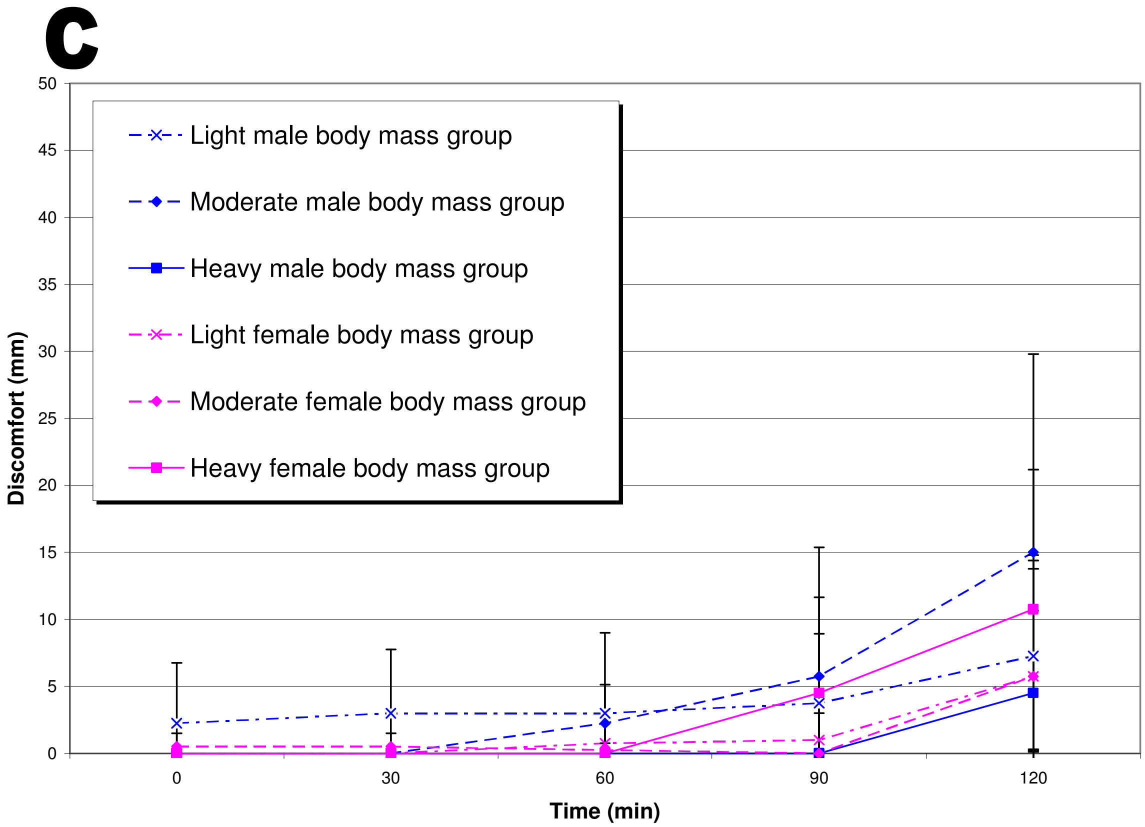
**Appendix C:** Left buttocks region discomfort trends. Trends are broken into male and female populations (A), light, moderate and heavy body mass groups (B), and light male/female, moderate male/female, and heavy male/female body mass groups (C).

A

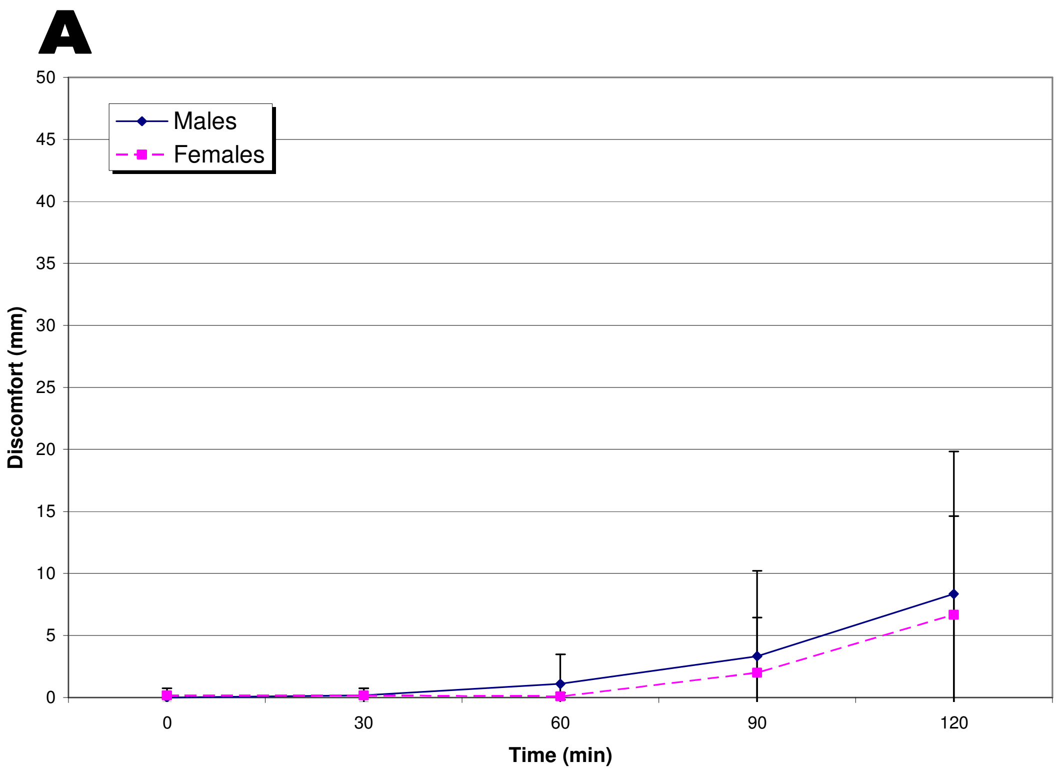


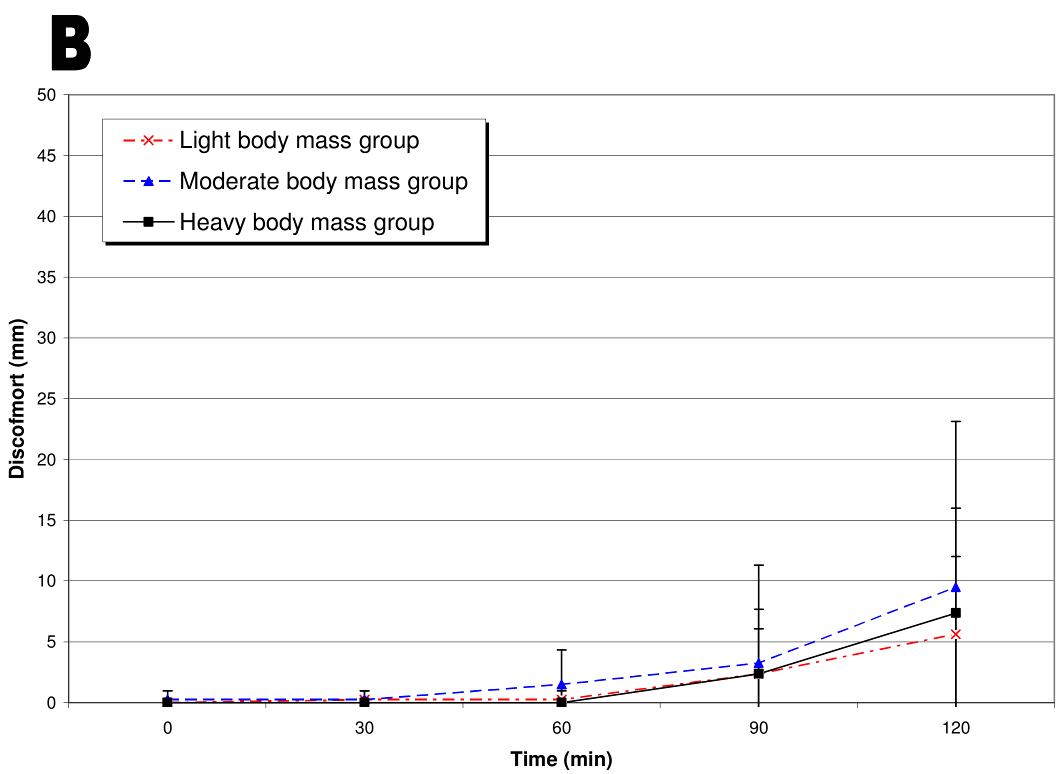

B

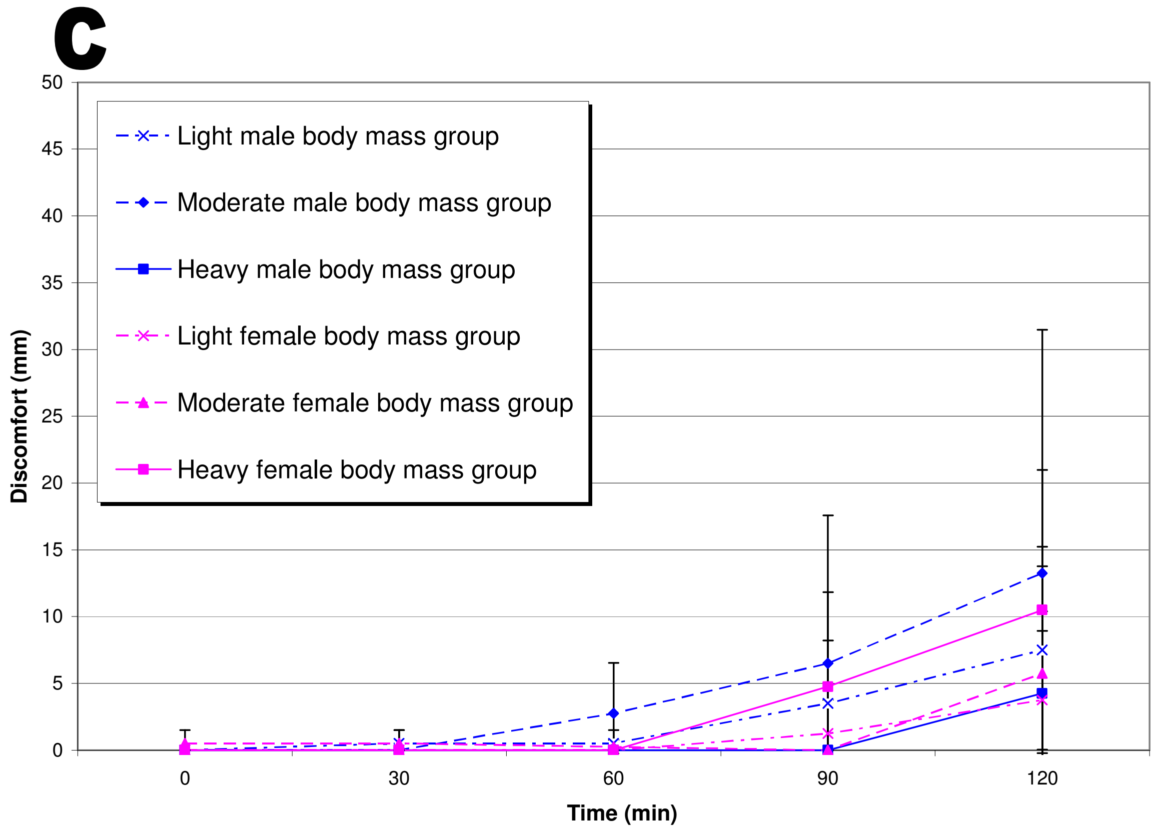



c
50
45
40
35
30
25
20
15
10
5
0
Discomfort (mm)
Light male body mass group
Moderate male body mass group
Heavy male body mass group
Light female body mass group
Moderate female body mass group
Heavy female body mass group
0
30
60
90
120
Time (min)

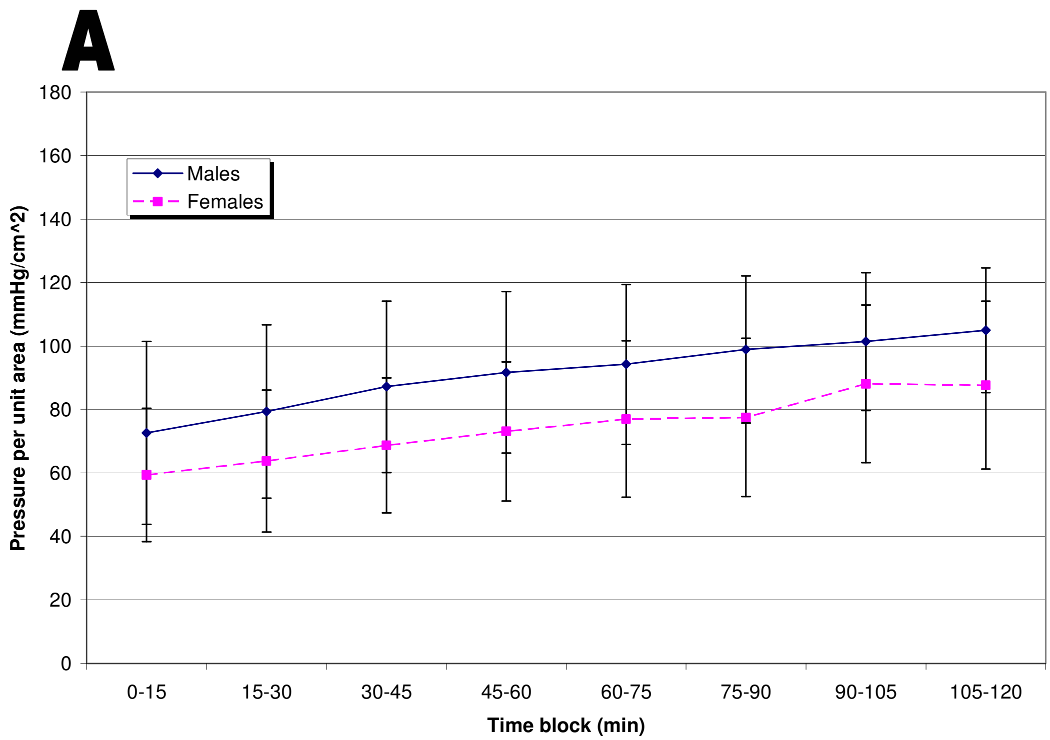
**Appendix D:** Right buttocks region discomfort trends. Trends are broken into male and female populations (A), light, moderate and heavy body mass groups (B), and light male/female, moderate male/female, and heavy male/female body mass groups (C).

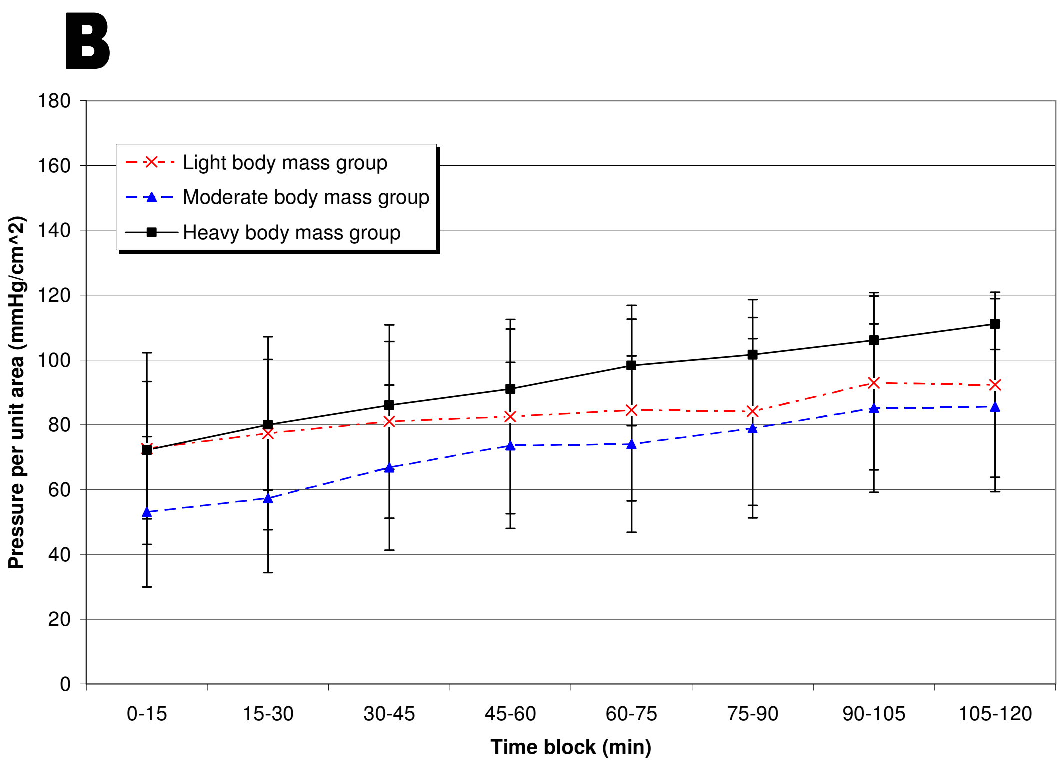


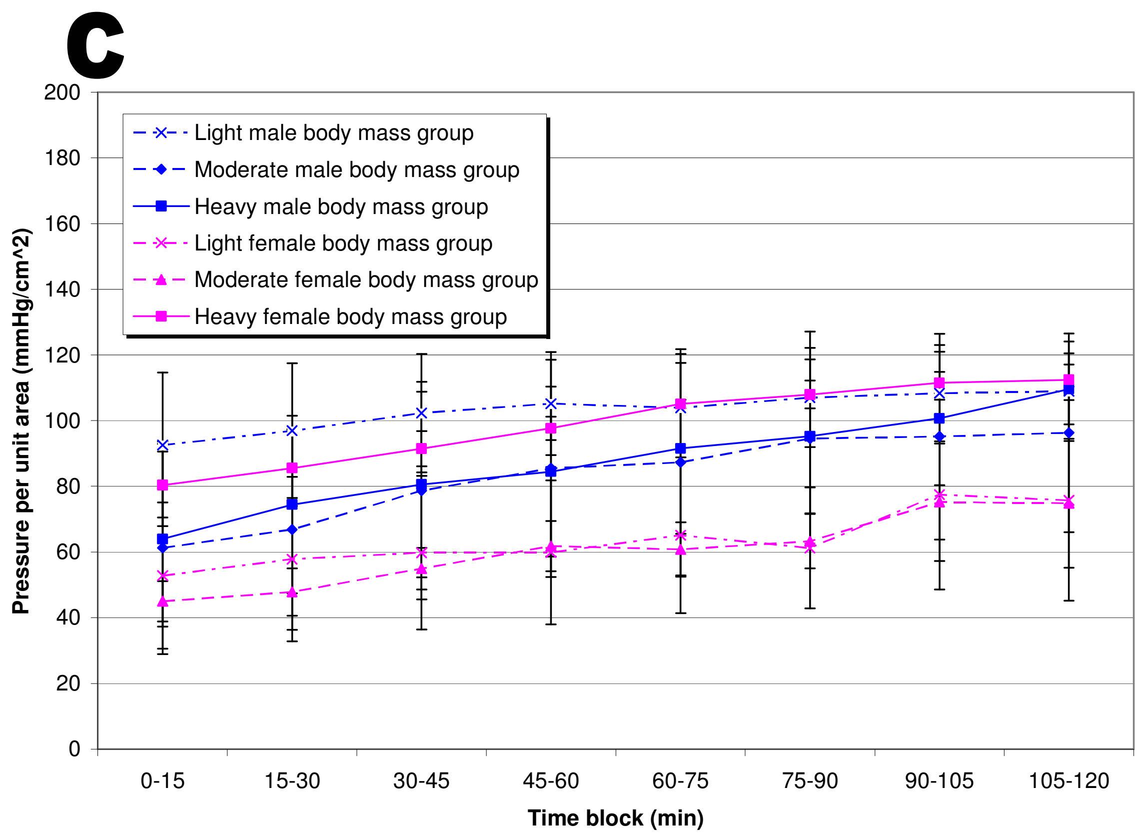




C
Light male body mass group
Moderate male body mass group
Heavy male body mass group
Light female body mass group
Moderate female body mass group
Heavy female body mass group
Discomfort (mm)
Time (min)
0
5
10
15
20
25
30
35
40
45
50
0
30
60
90
120

**Appendix E:** Time varying response of right IT PPA. Trends are broken into male and female populations (A), light, moderate and heavy body mass groups (B), and light male/female, moderate male/female, and heavy male/female body mass groups (C).

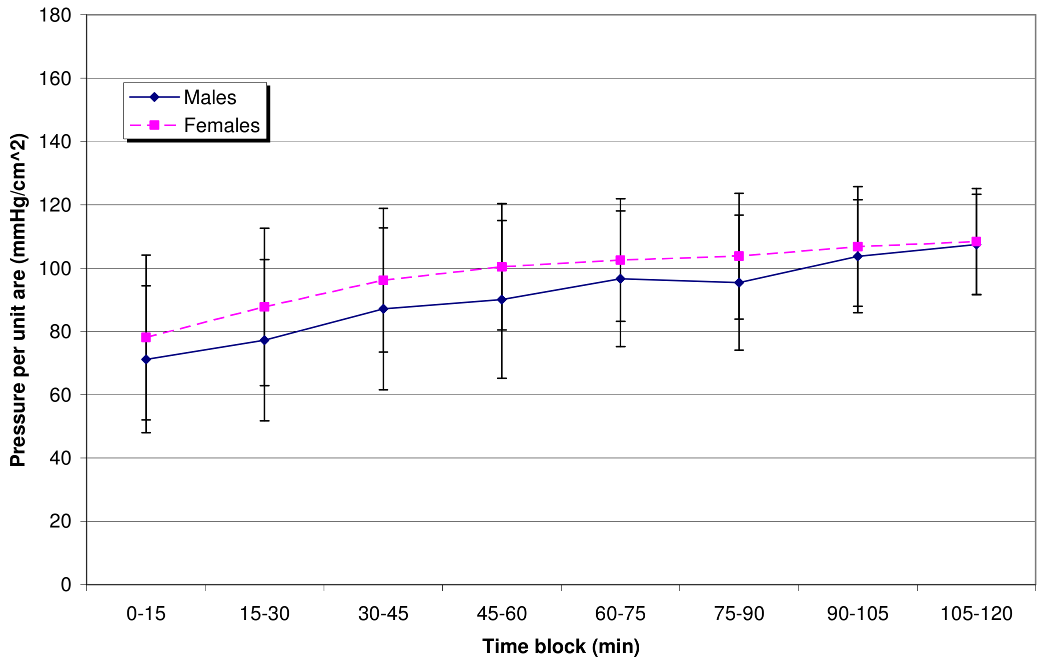


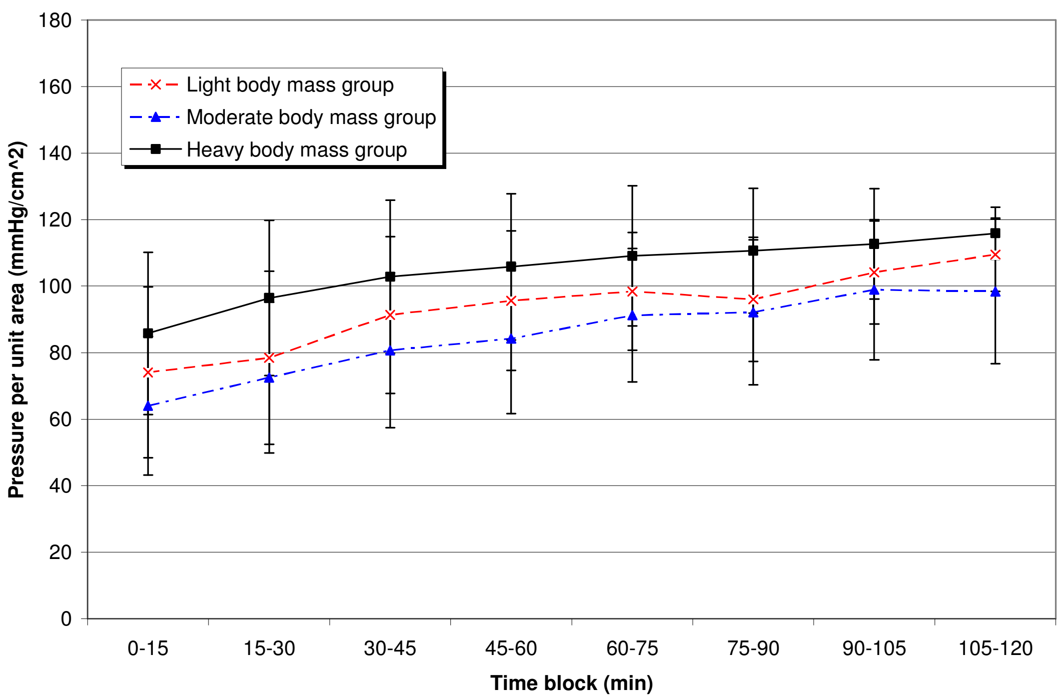

c
Light male body mass group
Moderate male body mass group
Heavy male body mass group
Light female body mass group
Moderate female body mass group
Heavy female body mass group
Pressure per unit area (mmHg/cm^2)
200
180
160
140
120
100
80
60
40
20
0
0-15
15-30
30-45
45-60
60-75
75-90
90-105
105-120
Time block (min)

**Appendix F:** Time varying response of left IT PPA. Trends are broken into male and female populations (A), light, moderate and heavy body mass groups (B), and light male/female, moderate male/female, and heavy male/female body mass groups (C).

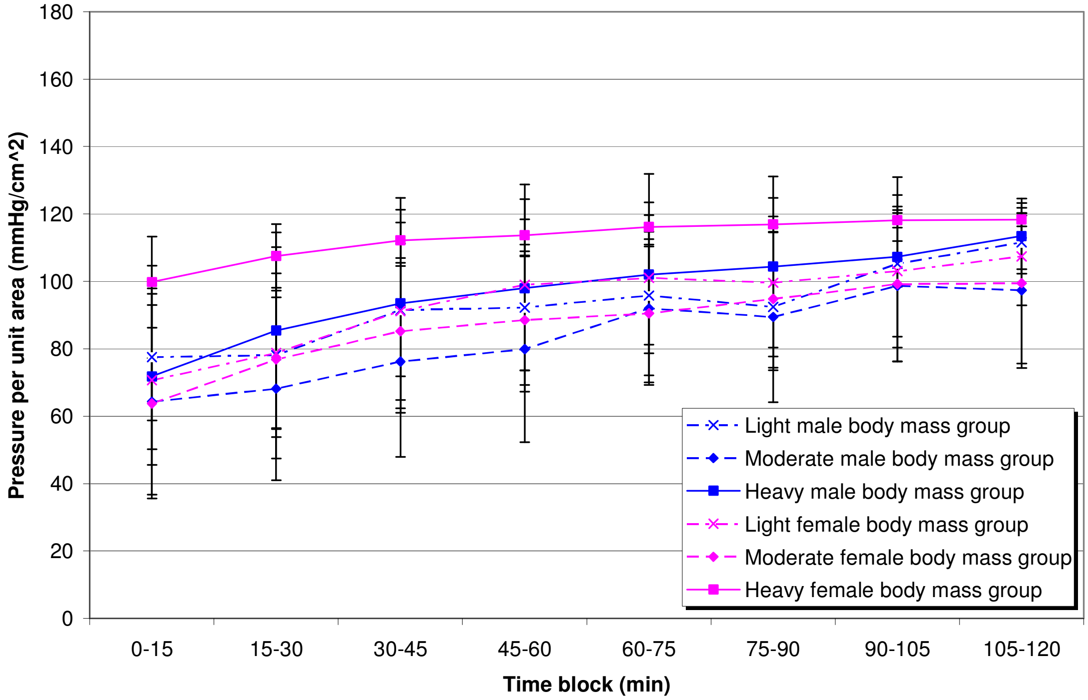
A




B



C



Pressure per unit area (mmHg/cm^2)
180
160
140
120
100
80
60
40
20
0
0-15
15-30
30-45
45-60
60-75
75-90
90-105
105-120
Time block (min)
Light male body mass group
Moderate male body mass group
Heavy male body mass group
Light female body mass group
Moderate female body mass group
Heavy female body mass group