Influence of a Marching Snare Drum System on
Joint Kinematics, Electromyography, and Contact Pressure

by

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ABSTRACT

Each year thousands of students participate in marching band. Many participate by carrying a flag, baton, woodwind or brass instrument, or drum. The drumline is of particular interest due to the unique and restricting nature of the instrument: a load carried anterior to the body with the only points of contact on the body at the shoulders and across the abdominal area. The influence of load carriage research on gait and joint mechanics, muscle activity, and contact pressure can be seen in the ever changing backpack designs and recommendations of load mass; however, there still remains a lack of research on marching band load carriage. The purposes of this investigation were: 1) to determine the influence of a standard marching snare drum system (SDS) and a modified marching snare drum system (MDS) on trunk and lower extremity kinematics while standing and marching; 2) to determine the muscular demand placed on the erector spinae, of individuals wearing a SDS and MDS while standing and marching; and 3) to determine the contact pressures at the shoulders and abdominal region of individuals wearing the SDS and MDS systems.

The results showed that the SDS during stationary tasks decreased hip flexion, and increased trunk flexion, ankle plantarflexion, ES EMG_{AVG} and EMG_{PEAK} as a result of the changes in posture, and contact pressures at the shoulders. When the participants wore the SDS while marching, the results showed increased trunk flexion, and increased ES EMG_{AVG} and EMG_{PEAK} as a result of the changes in trunk angle. When implementing

a lumbar belt there were similar significant findings at the joints and muscle activity level from the unloaded condition only, suggesting that the implementation of the lumbar belt did not alter the marching kinematics of the performer. The MDS condition however, did result in lower contact pressure values at the shoulders and higher pressures at the abdominal region, indicating a redistribution of pressure and load to the belt, hips, abdomen, and pelvic girdle and off of the shoulders. Therefore it can be concluded that implementing a lumbar belt to a SDS system ameliorates skin contact pressure and discomfort at the shoulders, potentially decreasing the occurrence of an injury, and anecdotally providing relief to the participants.

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List of Abbreviations

AA_{AVG}	Average Ankle Angle
ABS_{PEAK}	Peak contact pressure at the abdominal/pelvis area
BOS	Base of Support
COG	Center of Gravity
COG_{person}	Center of Gravity of an individual
COG_{pack}	Center of Gravity of the load carriage device
COG_{sys}	Center of Gravity of the person + pack
ES	Erector Spinae
HA_{AVG}	Average Hip Angle
KA_{AVG}	Average Knee Angle
LS_{PEAK}	Left Shoulder Peak contact pressure
MDS	Modified Marching Snare Drum System-Drum + Harness + Lumbar Belt
RS_{PEAK}	Right Shoulder Peak contact pressure
$SHLDR_{PEAK}$	Peak contact pressure at the shoulders
sEMG	Surface electromyography
SDS	Standard Marching Snare Drum System-Drum + Harness
TA_{AVG}	Average Trunk Angle

CHAPTER I

INTRODUCTION

Music is a universal language that can evoke every emotion, regardless if it resonates from a church choir, a national orchestra, or a college marching band. In order to achieve the desired performance, musicians must train. Train as a musician to perform the music and, in the case of marching bands, train as an athlete to perform the on-field performance and while carrying an instrument.

Marching music presentations require not only musicianship but also demanding movements, postures, and instrument weight carriage. This can provide a challenge for some of the larger awkward instruments or those whose carriage position alters the location of the center of gravity of the body. It is the load carriage demands of marching music presentations that this study will investigate. Specifically, the biomechanical effects associated with wearing a snare drum carrier and drum, and how the implementation of a lumbar belt to the snare drum carrier influences these effects.

There are many load carriage systems available on the market today such as backpacks, carrying bags, rucksacks, holsters, and belt clips. Some of these systems require that the load be carried on only one shoulder, supported by both shoulders, or supported on the pelvis and hips. The present study will investigate the dual shoulder method and implementing a device to place a portion of the load on the pelvis, and as such will not address the issues associated with asymmetrical loading.

Load Carriage

Load carriage allows an individual to transport an additional mass whether it is on the anterior part, posterior part, or sides of the body, or through the use of a carrying device (i.e., backpack, rucksack, or carrier). Examples of load carriage that are found in the research include: hiking backpacks (Cook & Neumann, 1987; Goh, Thambyah, & Bose, 1998), school bookbags (Devroey, Jonkers, Becker, Lenaerts, & Spaepen, 2007), military load bearing equipment (Ling, Houston, Tsai, Chui, & Kirk, 2004; Martin & Nelson, 1986), and ergonomic/lifting (Anderson, Meador, McClure, Makrozahopoulos, Brooks, & Mirka, 2007; Cham & Redfern, 2004). However, even with the abundance of individuals participating in marching bands throughout the country, there is little research on the influence of load carriage in marching bands. Nearly absent are investigations into the anterior load of a standard marching snare drum system (SDS-snare drum and carrier). The few studies available on marching bands have only examined whether or not participation in that activity can be a substitute for a physical education class (Strand & Sommer, 2005) or examined a survey on marching band injuries sustained during a season (Mehler, Brink, Eickmeier, Hesse, & McGuire, 1996). The characteristics of the SDS are unique and include: the carrier has minimal adjustment capabilities; the mass of the snare drum will always be located anteriorly to the individual; there is a lack of a hip or back support on the carrier; and there are only three points of contact throughout the entire carrier (upper trapezius region, upper back-scapular region, and an abdominal plate). Research on backpacks, rucksacks, and ergonomic tasks provide the foundation for understanding the biomechanical influence of load carriage systems on the human body, and for that reason, understanding the impact of the SDS and beginning to establish this line of research will be inferred from research on posterior and anterior load carriage.

Biomechanical Effects of Load Carriage Mass

Posterior & Combined Anterior/Posterior Load Carriage

One of the most popular methods of load carriage in the current era is the bookbag. It is estimated that over 40 million individuals, primarily students, use bookbags as a means to carry personal belongings, work-related materials, and particularly school supplies and books. It is recommended that the maximum about of load carriage placed on the body should not exceed 20% of body mass (BM) (Malhotra & Sen Gupta, 1965; Pascoe & Pascoe, 1999). The student population however, can end up carrying loads weighing more than the 20% of the individuals' BM from the textbooks alone, before additional supplies, lunches, and other miscellaneous materials are calculated into the total load. It is these large loads that are believed to contribute to the significant kinematic and kinetic changes that have been associated with back pain or injury for those who utilize this method of load carriage (Devroey et al., 2007; Hong, Li, & Fong, 2008; Malhotra & Sen Gupta, 1965; Pascoe & Pascoe, 1999).

Another popular method of load carriage, and one that is as specific to a particular activity is a hiking pack. Hiking packs are unique from bookbags by utilizing either an internal or external frame and often use lumbar belts which provide the pack with an additional stabilizing and load absorption component, (Reid, Stevenson, & Whiteside, 2004; Smith, Ashton, Bohl, Clark, Metheny, & Klassen, 2006). The literature provides evidence that the addition of support structures (lumbar belt and/or frames) aid in the transfer of forces from the body to the frames (Reid et al., 2004), shift the center of gravity of the system (COG_{sys}) location of the load closer to the body (Bloom & Woodhull-McNeal, 1987; Goh et al., 1998), transfer forces from one part of the body to another more suited for load carriage (i.e., from the shoulders to the pelvis and hips), and decrease the muscular demand to support the load (Holewijn, 1990; Harman, Han, Frykman, Johnson, Russell, & Rosenstein, 1992).

Military rucksacks are examples of a posterior load carriage system, as are hiking packs and school backpacks and therefore cause individuals to demonstrate similar responses. Military rucksacks utilize rigid external frames, similar to hiking packs, which allow for a more stable load carriage system for the soldiers (Bloom & Woodhull-McNeal, 1987). The difference between the load carriage systems is that a soldier does not have the option of managing the items carried and therefore does not have control over the weight of the pack. While the external frames are light weight, they still contribute to the overall mass of system in addition to the mission-required equipment. A military rucksack has a mass of approximately 21 kg empty; therefore with the addition of the required supplies the mass of the system can reach well over the recommended 10-20% BM that should be placed on the body (BAE Systems, 2010; Devroey et al., 2007; Hong, Li, & Fong, 2008; Malhotra & Sen Gupta, 1965; Pascoe & Pascoe, 1999). Military load carriage can also consist of loads being placed on the anterior and posterior aspects of the body, including the utility vest worn under the rucksack , body armor, and double packs (Birrell & Haslam, 2009; Harman et al., 1994; Kinoshita, 1985). Research assessing the effect of the double packs (half the mass anterior, half the mass posterior) revealed similar kinematic results to anterior and posterior load carriage methods, specifically; decreases in stride length and an increase in stride frequency (Knapik, Reynolds, & Harman, 2004).

An individual's body alignment adapts to increased loads as a means of alleviating the impact of the load. The trunk adopts a more forward leaning position to accommodate the posterior shift of the individual's center of gravity (COG_{person}); a forward neck position is adopted to assist the increased trunk flexion in shifting the COG_{sys} forward, there is decreased balance control, and therefore may lead to subsequent lower back pain (Al-Khabbaz, Shimada, & Hasegawa, 2008; Devroey et al., 2007; Goh et al., 1998; Pascoe, Pascoe, Wang, Shim, & Kim, 1997). Furthermore, gait analyses reveal that increasing the mass of the backpack results in decreased stride length and single support time, and increased stride frequency and double support time (Kinoshita, 1985; Wang, Pascoe, & Weimar, 2001). It should be noted that there the center of gravity can be defined as the COG of the load/pack (COG_{pack}), the COG of the person (COG_{person}), and the COG of the pack/person system (COG_{sys}).

The static and dynamic responses to posterior and combined anterior/posterior load carriage include a posterior and vertical shift in COG_{sys}, the trunk adopts a more forward lean, there is increased double-support time, muscular activity, vertical and horizontal force distribution on the body, and subsequent loading of the lower extremities (Birrell and Haslam, 2009, 2010; Goh et al., 1998; Grimmer, Dansie, Milanese, Pirunsan, & Trott, 2002; Ling et al., 2004). Although carrying loads on the back has been the primary load carriage method throughout history (Knapik, Harman & Reynolds, 1996), there are other methods that allow an individual to transport items, for example: loads placed only on the anterior portion of the body, identical to the demands seen in marching drumlines.

Anterior Load Carriage

When carrying an external (i.e., a rifle, front packs, or office crates) or internal (i.e., pregnancy) load anteriorly, the static posture goals remain the same: perform the required task while limiting the amount of COG_{sys} excursion from that of an unloaded location, limiting excess muscular demand, and decreasing forces exerted on the lower extremities and within the vertebral column (Legg & Mahanty, 1985). In addition to consistency in static posture goals, many of the same outcomes noted during the gait cycle in posterior load carriage hold true for anteriorly positioned loads. These findings include a more abrupt loading phase during gait, a shift in the COG_{sys} vertically and anteriorly, and an increase in hip and knee flexion angle (Anderson et al., 2007; Cham & Redfern, 2004).

Pregnancy research supports these findings and has shown that the anterior shift of the COG_{person} in addition to the increase in body mass contributes to an increase in anterior pelvic tilt and double support time, and decreases in hip extension and single support time (Foti, Davids, & Bagley, 2000). The result of those kinematic adjustments require an increase in muscular demand, and may provide an explanation of the musculoskeletal/overuse issues seen in anterior load carriage. Myung and Smith (1997) investigated anterior load carriage and indicated that stride length was influenced by the mass of the load; however, the load investigated was 40% BM, well above previous research testing loads. Although the research to suggest a relationship between anterior load mass and stride length were studied at much lower values (20% BM), the overall findings in this research are indicative of the real-world and therefore should be taken into account.

Biomechanical Effects of Vertical Load Carriage Location

In addition to the load carriage mass, research on load carriage includes investigating the result of the load carriage's vertical position. It is recommended that the load be positioned as close to the COG_{person}, in both the vertical and horizontal directions. Increasing the distance of the load away from the COG_{person} in any direction can compromise an individual's balance (Qu & Nussbaum, 2009) and increase strain on the body. Carrying a load just 30 cm horizontally from the hip, for example, has been shown to cause a trunk flexion moment of up to 20 Nm. The result is an equal but opposite extension moment to counteract the forward tilt, and increased muscular demand (Cham & Redfern, 2004).

Higher load placement during static conditions may allow an individual to shift the COG_{sys} more easily in an attempt to maintain balance; however, in order to maintain that balanced state, there is an increase in the muscular demand due to the larger moment that is created. Maintaining a higher COG_{sys} also means a decrease in stability (less allowable movement within the base of support (BOS)), which actually helps initiate movement and is therefore preferred for level walking. Lower placement for static conditions is thought to cause more adjustments by the body to maintain balance, yet a lower COG_{sys} is more stable and can result in less muscular demand due to a smaller moment arm. A lower COG_{sys} resulting in better balance during movement has led other investigators to believe it is better for walking on uneven terrain.

Southard and Mirka (2007) researched dynamic conditions and recommend lumbar height placements during level walking conditions for posterior loads to maintain balance and decrease muscular demand. Bobet and Norman (1984) agree and recommended placing the load in the lower thoracic region during uneven terrain movements as high placements can result in increased moments and muscle activity at the lower back. Knapik et al. (2004) concur that lower load placement is better suited for uneven terrain by lowering the COG_{sys} closer to the ground, and higher load placements are better suited for level walking as it helps initiate the movement. The conclusion, vertical load placement recommendations are complicated and unclear.

Load placement height issues differ between anterior/posterior loads, static/dynamic movements, even/uneven terrain, and research in this area is less abundant that the literature on load carriage mass. Regardless, the method of carrying (anterior versus posterior) load mass (% body mass) and location (thoracic versus lumbar regions) are all factors affecting gait, stability measures, and muscular adaptations.

Muscle Activity & Contact Pressure

Load carriage will predominately affect the primary point(s) of contact, whether the load is placed on the legs or arms (Knapik et al., 2004), or for the purposes of this study, the shoulders. In addition to affecting the kinematics of the individual during stationary and dynamic tasks, consideration must be given to the outcomes of muscle activity and skin contact pressures due to load carriage.

Surface Electromyography

Any deviation from the normal, unloaded, anatomically neutral posture results in a change in muscular demand (Bobet & Norman, 1984; Hamilton, Weimar, & Luttgens, 2008). Deviations that occur during load carriage situations can be from increased load mass, changes in the amount and location of force being exerted on the body, changes in load placement, or a combination of any of these factors. Researchers not only found that increasing load mass causes an increase in muscular activity, but also supported the recommendation that loads not exceed 15% of body mass (Al-Khabbaz et al., 2008; Devroey et al., 2007). In addition to investigating the changes in muscle activity due to load carriage mass and location, researchers also considered the length of time that the load was being carried. The results indicated that prolonged load carriage can lead to the excess muscular tension that can potentially lead to muscular pain and fatigue (Bobet & Norman, 1984; Hong et al., 2008).

While gait and balance demands associated with load carriage involve the whole body, this discussion will be limited to the muscles identified in the research as being most involved with posterior and anterior load carriage: i.e., erector spinae (ES), rectus abdominis (RA), and upper trapezius (UT) (Al-Khabbaz et al., 2008; Bobet & Norman, 1984; Devroey et al., 2007; Hong et al., 2008; Schiffman, Bensel, Hasselquist, Gregorczyk, & Piscitelle, 2006). Bobet and Norman (1984) investigated the difference between the trapezius (TRAP) and ES for loads carried at the ear lobe level (first cervical vertebrae) and xiphoid process (tenth thoracic vertebrae). These researchers found the midback load placement resulted in less muscle activity in both the ES and TRAP muscles, with the erector spinae muscles exhibiting less activity overall than the TRAP

(Bobet & Norman, 1984). Al-Khabbaz et al. (2008) found that the RA activity, but not the ES, changed significantly at loads of 10% BM or greater in backpacks. This occurred due to the posterior load on the trunk causing an increase in trunk extension. Although the ES are trunk stabilizers, the significant muscle activity occurred in the trunk flexors in order to counteract the extension moment. Hong et al. (2008) found that backpacks weighing 15-20% BM significantly increased the muscular demand of the lower TRAP due to the increased demand on the shoulders from the backpacks.

Load carriage has also been found to increase trunk stiffness (increased level of muscular contraction). Muscular stiffness can occur as a result of the joint(s) being hindered in movement (Birrell & Haslam, 2008) or due to a decrease in stability as a result of an external load (Shape, Holt, Saltzman, & Wagenaar, 2008). During load carriage situations, a decrease in stability of the trunk can occur due to the additional flexion, extension, lateral, and rotational moments that occur as a result of the load carriage system shifting the COG_{person} away from the unloaded position. The result is an increase demand placed on the musculature that can lead to increased pain, early onset of fatigue, and injuries.

Load mass and placement has been shown to influence muscle activity. Increased muscle activity occurs due to the body being loaded as well as the additional adjustments that are made as the body attempts to compensate for the extra load. For example, the responses occur in the musculature opposite to the joint action (i.e., increased abdominal activity due to the trunk flexion as a response to spine extension caused by the load). Research to date has centered on the points of the body with which the load makes contact (i.e., the trunk), instead of the moments that are created as a compensatory mechanism. The research on the muscle activity responses of the lower extremities to load carriage is limited, but the trends found by Al-Khabbaz et al. (2008) indicate that the lower extremity musculature will show increased activity in an assisted role. Therefore, more research is needed to determine whether load carriage influences the lower extremity muscles (i.e., gastrocnemius and soleus), or whether the trunk musculature is solely responsible for managing the increased demands.

Contact Pressure

In addition to investigating the kinematics and muscle activity associated with load carriage, skin contact pressure has also been examined as it has been considered a limiting factor during load carriage (Holewijn, 1990).   Increased load mass impacts not only the kinematic responses of the body but also causes an increase in skin contact pressure (Holewijn, 2000; Mackie, Stevenson, Reid, & Legg, 2005) and can have deleterious consequences on the individual. Skin contact pressure of more than 105mm Hg (14kPa) results in irritation and redness; and if exposed for more than two hours, may lead to subcutaneous edemas (Husain, 1953). High pressures are also indicative of the impact the load has on the structures below the surface of the skin, for example the superficial nerves of the brachial plexus. If these structures are pinched, damaged, or compressed, the result is a condition known as Rucksack Palsy. This pressure decreases the efficacy of the neuromuscular signal from reaching the target muscle resulting in weakness and/or loss of function (Goodson, 1981; Knapik et al., 1996, 2004).

In an effort to combat the effects of load carriage not only on the joints, muscle activity and contact pressure, researchers have investigated the implementation of a lumbar belt. In a study particularly related to skin contact pressure, the researcher noted a 36% increase in pressure as the load mass increased within the pack system without a lumbar belt (Holewijn, 1990). This study was able to show a relationship between load mass and load carriage type, and the significance of a lumbar belt.

Lumbar belt Implementation

In addition to suggesting lighter masses and alternate loading positions, there is also research investigating the impact of implementing supportive devices, such as a lumbar belt. LaFiandra and Harman (2004) investigated the forces exerted on the upper and lower back and found that the addition of a lumbar belt helped redistribute the location of the forces on the body to the pelvis. Redistributing these forces to the pelvis is beneficial because the hips are three times less sensitive to pressure, are located closer to the COG_{person} position, and are the preferred (anecdotally) placement for load carriage (Knapik et al., 2004; Scribano, 1970). Sharpe et al. (2008), expanding on the work by LaFiandra, Wagenaar, Holt, and Obusek (2003), examined the effect of implementing a lumbar belt on the coordination between the pelvis and trunk movements and found that the belt improved the stability of the pelvic-thorax coordination pattern by decreasing the muscular demand of the trunk stabilizers. Essentially, the lumbar belt allowed for less restrictive pelvic-trunk rotation, while providing stability between the segments and thereby decreasing the muscular demand.

Implementing a lumbar belt has been shown to be beneficial in the relationship between muscle activity and skin contact pressures. Holewijn (1990) for example found that 10kg carried in a frameless pack resulted in a peak skin pressure of 203mmHg at the shoulders; that same mass carried in a pack with a flexible frame and lumbar belt resulted in decreased TRAP muscle activity and a peak pressure of only 15mmHg. While Holewijn did not ascribe the decrease in muscular activity to the lumbar belt alone, a subsequent study showed that without the use of a lumbar belt, there was a 44-47% increase in shoulder pressure (Mackie et al., 2005).

Loads produce tension on the anterior and superior aspect of backpack shoulder straps and thereby contribute to the increased pressure exerted on the shoulders. The amount of tension and pressure is influenced by load mass, load location, tightness of straps, and use or non-use of a lumbar belt. Without the use of lumbar belt, both the skin contact pressure and increased muscle activity become significant and potentially task limiting factors. With the aid of a lumbar belt a portion of the load can be redistributed to the waist, decreasing the pressure and muscular demands of the shoulders and providing for a more efficient method of load carriage.

Summary

Load carriage has many ramifications on the body such as: decreased balance and stability (shifting the COG_{person} more to the edge of the BOS), increased muscular demand, and increased potential for injury. In particular, research has demonstrated that load carriage decreases stride length (Birrell & Haslam, 2009; LaFiandra et al., 2003), single support time (Attwells, Birrell, Hooper, & Mansfield, 2007; Wang et al., 2001), hip extension angle (Anderson et al., 2007; Cham & Redfern, 2004), and increases double support time (Goh et al., 1998; Kinoshita, 1985; Wang et al., 2001), stride frequency (LaFiandra et al., 2003; Martin & Nelson, 1986), and knee flexion angle

(Anderson et al., 2007; Devroey et al., 2007; Holt et al, 2003; Kinoshita, 1985). Overall, these adaptations were seen during bookbag, military, hiking, and ergonomic research, all areas where there are fewer performance constraints on the individual. Although there are some similarities between military and marching band load carriage (i.e. regimented step lengths and cadences), and yet there is a dearth of research within the marching arena. Marching music performers are also constrained by the body posture, step frequency, and having a fixed load. These limitations highlight the inability to automatically apply the solutions found in load carriage research to marching load carriage tasks.

The findings of the noted research are expected to be confirmed in the present study if the marching snare drummer were not to be constrained in a number of ways. Specifically: (a) there is no horizontal adjustment on the SDS; (b) The vertical location of the drum is set for the proper playing position first and the aesthetics of the drum line second. The height of the individual has little influence on the height of the drum; (c) There is a standardized step size, no matter the leg length of the individual (and often a standardized stride frequency); (d) The rigidity of the carrier results in higher contact areas as the shape of the carrier cannot be adjusted, and (e) the position of the drum (directly anterior to the thighs limiting hip and knee angles) prevents this population from employing the protective and stability mechanisms seen in the more common load carriage situations.

Statement of the Problem

Currently there are thousands of individuals participating in marching bands, yet there is little research on the topic. Within the literature on marching bands, the most frequently discussed topic is whether or not marching band can be considered a replacement for physical education classes (Cowen, 2006; Erdmann, Graham, Radlo, & Knepler, 2003; Strand & Sommer, 2005). The medical issues encountered throughout a marching season have also been investigated (Jones, 1983; Mehler et al., 1996) but still remains a topic less frequently investigated. In order to advance the research in the area of marching band load carriage, the first step will be to determine if the findings of the proposed study are concurrent with previous literature on load carriage. The next step will be to investigate the influence of the addition of a hip support to the SDS, and the role of the lumbar belt in alleviating the negative side effects anecdotally seen in marching band load carriage.

Statement of Purpose

The purposes of this research are: 1) to determine the influence of a SDS on trunk and lower extremity kinematics while standing and marching; 2) to determine the contact pressures and muscular demand placed on the erector spinae, of individuals wearing a SDS; and 3) to investigate the influence of a lumbar belt (MDS-modified SDS) on joint kinematics, muscular activity, and contact pressures. A standard snare drum system (SDS) consists of the initial carrier (**Figure 1**) and drum (**Figure 2**), and an MDS consists of the SDS with the addition of a lumbar belt.

Hypotheses

Primary Objective-To determine the postural demands of a marching snare drum system

1) Evaluate the joint kinematics during a load-carrying task using both an SDS and MDS in marching drumline individuals.

H_{01}: During the stationary movement condition, the SDS will result in increased trunk extension (smaller trunk flexion angle), hip flexion, knee flexion and plantarflexion.

H_{02}: During the dynamic movement conditions, the SDS will result in increased trunk extension, decreased hip flexion, increased knee flexion and increased dorsiflexion.

Secondary Objective-To determine the muscle activity and contact pressures associated with wearing a marching snare drum system

1) Evaluate the muscle activity of the erector spinae during a load-carrying task using both an SDS and MDS in marching drumline individuals.

H_{01}: The participants will demonstrate higher levels of muscle activity while wearing the SDS, during both the stationary and dynamic movement conditions.

2) Evaluate the pressures at the shoulders and abdominal/pelvic region of marching band individuals during a load-carrying task using both the SDS and MDS during the stationary movement condition.

H_{01}: The participants will demonstrate higher contact pressures at both the shoulder and abdominal region from the SDS.

H_{02}: The participants will demonstrate reduced contact pressures at the shoulders and increased contact pressures at the abdominal region from the MDS condition.

Limitations

The limitations for the present study include:

1. Self-reported health status was used.
2. The current marching snare drum system used by Auburn University Marching Band Snare Line consisted of a Pearl® Championship Maple FFX-1412 Snare Drum (**Figure 1**) with a Randall May® Aluminum Tube Snare Carrier (**Figure 2**).
3. The data collection took place at the beginning of the participants' season, when the drums are being worn on a regular basis.

Delimitations

The delimitations for the present study include:

1. Participants were required to wear retroreflective markers bilaterally and surface electrodes unilaterally.
2. Participants were between the ages of 19-25, members of the Auburn University Marching Band Drumline, and have at least one year (high school or college) experience performing with a marching snare or tenor drum.
3. The participants were assessed in a laboratory versus on the marching field or road (parade comparison).

4. The drum height was placed at an Auburn University standard, determined by the director, to be 10° below the horizontal for each individual.

Definition of Terms

Anatomical Joint Angle: The angle between two adjoining segments.

Base of Support: The area in which a part of the body is in contact with the supporting surface within the outline of both feet (Holbein & Chaffin, 1997).

Center of Gravity (COG): The point of a body about which all mass is said to act. The point of a body about which the system is balanced.

Kinematics: A branch of mechanics that investigates pure motion, without regard to the forces that produce that motion. Kinematic variables include position, velocity and acceleration.

Moment Arm: The perpendicular distance from the line of force to the point about which moments will be summed.

Plantarflexors: Muscles of the lower limbs documented to maintain upright stance as well as being involved in the propulsion phase of the gait cycle. Muscles included are the soleus, gastrocnemius, tibialis posterior, flexor digitorum longus, flexor hallucis longer, peroneus brevis and tertius.

Postural Stability: The body's capability to control the center of gravity within specific boundaries of space (limits of stability), enabling the individual to maintain in upright posture (Shumway-Cook & Woollacott, 1995).

Standard Marching Snare Drum System (SDS): The load carriage (LC) apparatus used for this study and by the Auburn University Marching Band Drumline include the Pearl® (Pearl® Corporation, Nashville, TN, USA) Championship Maple FFX-1412 Snare Drum (7.67kg) with a Randall May® (Randall May® International, Inc., Irvine, CA, USA) J-Rodless Contour Hinge™ Monoposto® Snare Carrier (0.91kg) **(Figures 1 & 2 respectively).**

Figures 1& 2. Snare Drum (Pearl, 2010) and Snare Drum Carrier (Randall May, 2010).

Surface Electromyography (sEMG): A technique for evaluating and recording the timing and magnitude of voluntary skeletal muscle activity. EMG is performed using an instrument called an electromyograph that produces an electromyogram (record of data) (MacIntosh et al., 2006; Winter, 1990).

CHAPTER II

REVIEW OF LITERATURE

Thousands of marching band musicians engage in demanding and strenuous movements regularly while practicing and performing, and a select group of musicians perform these movements while carrying large cumbersome loads. However, the influence of a marching snare drum system (SDS) on the biomechanics of a marching band member has received little attention in the literature. Therefore, this load carriage demand will be investigated in order to provide information about this population and the load carriage techniques being utilized. Specifically, this study will investigate how wearing a snare drum carrier and drum effects the trunk and lower extremities; and the influence of a lumbar belt on the biomechanical demands placed upon the bodies of marching band drummers.

This chapter is divided into five sections. Section one will summarize the previous work in posterior and combined anterior/posterior load carriage. Section two will describe the static and dynamic effects of anterior load carriage. Section three will describe the effects of load carriage on muscle activity and skin contact pressure. Section four will describe the impact of implementing a lumbar belt to a load carriage system. Finally, section five will summarize the pertinent findings of the previous literature to the present study.

Section 1: Biomechanical Effects of Posterior & Combined Anterior/Posterior Load Carriage

During unloaded conditions, the line of gravity of the head, arms, and trunk (HAT) is located anterior to the lumbosacral (LS) joint creating a more extension dominated posture (Bobet & Norman, 1984). This places an initial increased demand on the extensor musculature in order to maintain an erect posture and maintain the COG_{sys} within the base of support (BOS). It should be noted that there is the center of gravity of the load/pack (COG_{pack}), the COG of the person (COG_{person}), and the COG of the pack/person system (COG_{sys}). During posteriorly loaded conditions these adaptations are even more pronounced and include: increased muscle activity (Al-Khabbaz et al., 2008; Devroey et al., 2007), increased motion of the trunk and lower extremities (Chow, Leung, & Holmes, 2007; Devroey et al., 2007; Holt et al., 2003; Seven, Akalan, & Yucesoy, 2008), and a shift in the center of gravity (COG_{sys}) of the combined load and body system (Cook & Neumann, 1987; Holt et al., 2003). Likewise, posteriorly loaded gait trials also produce movement adaptations that include: decreases in single support time (SST) along with increases in double support time (DST) (Kinoshita, 1985), decreased stride lengths and increased stride frequencies (LaFiandra et al., 2003), changes in force distribution (Bloom & Woodhull, 1987; Reid et al., 2004), and subjective reports of pain and discomfort (Moore, White, & Moore, 2007).

During static posterior loaded conditions, although less movement is required to move the COG_{sys} within the BOS, a higher COG_{sys} is also less stable because of the longer moment arm in the sagittal plane, particularly during dynamic movements. However, depending on the proximity relative to the axis of rotation, a lower COG_{sys}

tends to require additional compensation by the body (i.e., movement at the trunk, hips, and ankles) (Bloom & Woodhull, 1987; Devroey et al., 2007). It is by understanding these principles of static load carriage and the possible outcomes from significant changes in posture that a foundation for understanding the effects of load carriage to dynamic situations can be provided.

Posterior loading during dynamic conditions alter the COG_{sys} excursion amounts and knee kinematics (Holt, Wagenaar, LaFiandra, Kubo, & Obusek, 2003), influences the forces measured at the shoulders and waist (LaFiandra & Harman, 2004; Mackie et al., 2005), and can also result in significant changes in joint kinematics (Harman. Frykman, Knapik, & Han, 1994; Kinoshita, 1985; Smith et al., 2006). Both Holt et al. (2003) and LaFiandra et al., (2003) discovered that during unloaded conditions when walking speed is increased, the body adopted a longer stride length, increased stride frequency, increased knee flexion, and increased transverse pelvis and trunk rotation in order to maintain the required speed. When the body was loaded however, walking and maintaining speeds greater than 1.0 m/s resulted in decreases in pelvic and thoracic rotation, decreased stride length, and an increase in stride frequency. The lack of movement between the pelvis and trunk is primarily due to increases in trunk and lower extremity stiffness, possibly from muscle co-activation in an effort to stabilize the pelvis and minimize the rotational forces on the already challenged upper body (Birrell & Haslam, 2009; Sharpe et al., 2008).

Similar to LaFiandra et al., (2003), Birrell and Haslam (2009) found that as the load mass increased, particular within a framed pack, the pelvic girdle rotation decreased. The decreased pelvic girdle rotation may have occurred not only as a protective mechanism described above, but also due to the restricted arm swing associated with carrying a rifle during one of the test conditions, (Birrell & Haslam, 2009). Natural arm movements modulate the excursions of the COG_{sys} (Birrell & Haslam 2008), therefore it can be assumed that limiting this movement, as seen during military load carriage with a rifle or a marching band snare drummer, can impede this response and cause increased demands elsewhere in the body and result in limited joint range of motion (ROM). Military load carriage is also unique in that it places a demand on all sides of the body for load carriage (anterior and posterior). This occurs in the load carriage vests worn under the rucksacks, the webbing system vests worn, canteen belts, and body armor, and is referred to as combination load carriage. One attempt to combat the decreased stability of the pelvis due to load carriage, while still allowing the joints to move in an efficient manner, is to implement a lumbar belt. This topic will be discussed further, but as was noted by Sharpe et al. (2008) that loading participants with a lumbar belt equipped backpack resulted in significantly more pelvic girdle-to-trunk rotation pattern in addition to a more stable pattern in the participants' gait.

Dynamic studies on combination load carriage have revealed similar kinematic results to separate anterior and posterior load carriage methods such as a decrease in stride length (Birrell & Haslam, 2009; Harman et al., 1994), decreased trunk flexion (Harman et al., 1994) and an increase in stride frequency predominately due to the body being externally loaded. Kinoshita (1985) found that as the load mass increased, regardless of a posterior load or combination load design, the period of DST increased, knee flexion increased, and SST decreased. The significant change in knee flexion aided in the shock absorption during the loading phase, and correlated with the increased DST which may facilitate in the transition of weight between the feet. Kinoshita (1985) found that as the load mass increased in the backpack the participants utilized increased step widths which is indicative of decreased pelvic girdle rotation and a decrease in balance. The doublepack system resulted in less forward trunk lean and less altered gait kinematics. This difference between the double pack and single backpack have some researchers (Birrell & Haslam, 2010; Kinoshita, 1985; Knapik et al., 1997; Motmans, Tomlow, & Vissers, 2006) considering full body load distribution rather than one sided loading as the best option.

Both males and females will be included in the present study; therefore a brief review of the gender differences during load carriage tasks is appropriate. A majority of the load carriage research uses males only as participants. Presumably because of the predominance of men in the military, and because most load carriage devices are built based on male anthropometry. However, according to the U.S. Census (2010), there has been an increase in the number of females in the military: 197,900 active duty in 2008 to 213,823 in 2010 (Department of Veterans Affairs, 2010; infoplease®, 2011). Females have structural differences that affect the ability to carry loads in the same manner as men. For example, females typically have a shorter stride length and must increase stride frequency during ruck marches to maintain the same speed as the longer striding males. Females also generally have less muscle mass, less bone density, and increased joint laxity during the menstrual cycle; all of which may contribute to muscular or skeletal injuries during a mission (Ling et al., 2004). Females also have a wider pelvis that is not well suited to the narrow lumbar belts found in most rucksacks or hiking packs. This mismatch can result in discomfort at the pelvis as well as excessive movement of the pack (Birrell & Haslam, 2009; Ling et al., 2004). Another difference between males and females that should be taken into consideration is the general smaller size of the female and thus the increased load mass relative to body mass that must be carried.

In response to these differences, Ling et al. (2004), and Martin and Nelson (1986) chose to investigate the kinematic and temporal responses of females to different load carriage situations. Ling et al. (2004) investigated the influence of a military load carriage system (vest and rucksack) of 0, 20, 30, 40, and 50 pounds on seven young-adult females. Each condition consisted of three, pre-march walking trials at a speed of 4.827 km/hr across a 40 foot walkway; and a 60 minute simulated march trial on a treadmill at 4.827 km/hr; followed by another three post-march walking trials across the walkway. Ling et al. (2004) noted that the female participants responded to the increased load similar to men, with increased DST, decreased SST, increased hip and trunk flexion, and increased vertical COG_{sys} movement. The females also displayed decreased knee flexion, thus increasing the stiffness of the lower leg at impact and provided less shock absorption. Although there were no significant changes in stride length or stride frequency, the result did indicate similar trends shorter stride lengths and faster stride frequencies seen in previous studies. The insignificance could have occurred as a consequence of the researcher's stringent significance level ($\alpha < 0.005$). It should be noted that one participant was unable to complete all the trials due to the substantial discomfort over the pelvic girdle from the pack. The issue was due to an anatomical difference of having a small waist and wide pelvis that could not be fitted comfortably with the pack. In addition, three participants had to make modifications (i.e., custom pads added to the points of contact on the pelvis) in order to sufficiently shift the load from the shoulders to the hips for better comfort. These results provide further evidence of the need to design a load carriage system that is more adaptable across the sexes and body sizes.

Martin and Nelson (1986) compared eleven male and eleven female ROTC students while walking over level ground at 1.78 m/s with the same absolute mass conditions (9, 17, 29, and 36 kg). The results revealed that both males and females had decreased stride lengths and increased stride frequencies as the load increased. Moreover, the females had overall shorter stride lengths and increased stride frequencies than the males for all conditions. Both groups had decreased swing times as the load increased, yet the females had significantly shorter swing times from the males for the heaviest loads (17, 29, and 36 kg). The females showed an increase in DST as the load mass increased, with significant changes from the males occurring for the two heaviest loads. The males demonstrated less consistent changes in the DST during the loading conditions, but did yield a significant difference for the 9 kg and 36 kg loads. Both the males and females had significant trunk flexion for the 29 and 36 kg loads, with the females maintaining larger trunk angles overall, suggesting no effect of sex. Overall, the females had less trunk lean than the males and significantly less lean during the first three conditions. It was hypothesized that less trunk lean may have occurred as a result of already having a lower COG_{person}, and therefore the females were less influenced by the increased rucksack load.

Research has indicated that placing an external load posteriorly on the body or on both the anterior and posterior side of the body result in similar coping strategies. Although the posterior and combined literature provide the foundation for studying marching drum load carriage, it also reveals the current gap in the marching band load carriage literature. For example, the constraints of the marching gait limit adaptations such as decreases in stride length (Harman et al., 1994), decreased single stance time (SST) (Hong & Li, 2005), decreased pelvic rotation (Devroey et al., 2007), increases in stride frequency (Harman et al., 1994), and increased double stance time (DST) (Hong & Li, 2005). However, as unique as marching band load carriage tasks are, the literature on primarily anterior load carriage provides the structure for the changes in joint kinematics at the trunk, hip, knee, and ankle (Goh et al., 1998; Harman et al., 1994; Shasmin, Abu-Osman, Razali, Usman, & Wan Abas, 2005), changes in knee angle to aid in shock absorption (Kinoshita, 1985), and increased muscle activity (Devroey et al., 2007) that are anticipated to occur during this study.

Section 2: Biomechanical Effects of Anterior Load Carriage

The kinematic and kinetic results of carrying a load anteriorly are similar to posterior and combined anterior/posterior load carriage. Specifically, the following alterations have been identified: a shift in the COG_{sys} vertically and anteriorly (instead of posteriorly), an increase in trunk extension (instead of flexion), and increases in hip and knee flexion angle (Cham & Redfern, 2004; Anderson et al., 2007). Although there is limited research regarding anterior load carriage, there is research in the field of ergonomics that can provide understanding of the effects of being loaded or carrying loads on the anterior portion of the body. Anderson et al. (2007) for example, investigated anterior load carriage on posture while eleven healthy college-aged males stood still for 6 seconds and walked at a self-selected pace (ranging from 1.9-3.3 mph) on a level treadmill for 15 seconds. The participants performed trials with two loads: a hand-held barbell, mass equaling 20% of participants' maximum voluntary isometric elbow flexion, and a 14kg bucket. Both objects were held in the front on the body at heights in line with the knuckles, elbows, and shoulders. Overall, the trunk ROM while walking was 1.5° more than standing at the ninth (T9) and twelfth thoracic (T12) and third lumbar vertebrae (L3). Holding the barbell at all three heights resulted in increased trunk extension at T9 and T12 vertebrae during both the standing and walking conditions. When standing with the 14kg bucket at shoulder height, trunk extension not only occurred for both T9 and T12 vertebrae, but the L3 vertebrae as well. Similar to the barbell condition, carrying the bucket at shoulder height resulted in the largest trunk extension angle during gait, with similar ROM differences. The results of this study support previous literature on posterior load carriage that suggests that the trunk will adopt a leaning posture when loaded. This lean will move the COG_{sys} closer to the center of the BOS in order to try and maintain a balanced, upright posture. During posterior load carriage an anterior lean is observed; in response to an anterior load, a more posterior lean is utilized. Recall that an unloaded body presents with an extended posture due to the natural anterior carrying position of the head and trunk, maintained by the isometric contractions of the tonic musculature. Therefore, an anteriorly loaded body will require the trunk to extend even farther, putting a greater demand on this musculature.

Both Cham and Redfern (2004) and Myung and Smith (1997) investigated the impact of anterior load carriage and the surface of the floor on gait kinematics. Myung and Smith (1997) specifically investigated stride length as well as heel velocity due to a correlation between certain heel velocities and increased potential to slips/falls. Ten college-aged males, walked at a self-selected pace around a 25 m circular walkway under five loading conditions (unloaded, 20% BM in a container, 40% BM in a container, 18 kg container "held against the body", and 24 kg container "held against the body") over four floor surfaces (oily vinyl tile, dry stainless steel, oily plywood, and dry ceramic tile). Heel contact during normal gait ranges from 10-20 cm/s and can increase up to 50.8 cm/s during a slip. The researchers discovered that the heel velocity for dry floors was slower than the oily floors regardless of the loads, and the stride lengths were longer for the dry floors indicating the participants had better balance on the dry floors. Slippery surfaces required individuals to adopt a shorter stride length in order to maintain as much contact with the floor as possible and to keep the feet as close to the COG_{person} as possible. These adaptations resulted in increased stride frequency during fixed speeds, which in turn related to a higher heel velocity. These findings confirmed previous research that larger loads shorten the stride length, but the predominate influence on the gait was the floor surface. At a fixed speed the decreased stride length allowed for quicker transfer of load to the adjacent supporting limb, thus, decreasing time in single support and resulting in an increase in balance.

Cham and Redfern (2004) recruited ten healthy males to walk up a ramp positioned at 0, 5, and 10°, during three loading conditions (no load, 2.3 kg, and 6.8 kg). The load conditions consisted of weights placed in an office crate held bilaterally at waist level (elbows flexed approximately 90°) at a self-determined horizontal distance from the body. The results indicated significant increases in plantarflexion, knee flexion, hip flexion, trunk extension, as well as decreases in stance duration, and heel velocity. The increased plantarflexion, hip and knee flexion were affected by the changes in ramp angle across all angles, but only between the 0 and 6.8 kg loading conditions. While the changes in stance duration decreased linearly with the increased load mass, these outcomes occurred to a greater extent during the 0° angle compared to the 10° ramp. This occurs because the incline causes the foot to make contact sooner, therefore contributing to the increased knee and hip angles in order to swing the leg through quicker than when on level ground. Cham and Redfern's (2004) study suggests that the changes in these kinematic variables may help an individual decrease the likelihood of slipping. The significance of which is because of the various surfaces that a marching band individual will come across throughout the season including hills on campus and wet parade routes. However, once the slip occurs chances of falling may increase due to the increased inertia of the person/load and by having the arms occupied with the load.

Research has shown similar to posterior load carriage, pregnant individuals have increased anterior pelvic girdle tilt and double support time, and a decrease in hip extension and single support time (Foti et al., 2000). Foti et al. (2000) tested fifteen healthy women during the third trimester of pregnancy and one year post partum. The participants were asked to perform repeated self-selected paced walks covering approximately 12 m on level ground. The kinematic changes found by Foti (2000) during the pregnant trials were believed to be compensations as the pregnant subjects attempted to maintain as non-pregnant a gait as possible. For example, pelvic width (width between the anterior superior iliac spines) and ankle separation increased during DST. Compensation included increases in the hip adduction moment in order to keep the feet more centered under the body thus avoiding wide BOS indicative of a waddling gait.

A larger ankle plantarflexion moment was used as a means of increasing the propulsive forces in order to maintain a forward/linear gait and overcome the increases in body mass.

Anterior loads cause an individual's balance and stability to be compromised as indicated by the changes in kinematic and kinetic responses (Abu Osman & Ghazali, 2002; Anderson et al., 2007; Cham & Redfern, 2004; Foti et al., 2000). One method of compensating for these changes is to increase flexion at the hip and knee, resulting in the ability to swing the leg forward to place the foot on the ground quicker. In doing so, the result is a longer stance phase, longer periods of DST, and a better transition of the load to the opposite limb (Kinoshita, 1985). Holbein and Redfern (1997) tested fifteen males while unloaded and carrying an 11.4 kg box during a functional limits of stability test. The researchers noted that increased flexion at the hip and knee lowers the COG_{sys} thereby increasing stability and extending an individual's stability limits. When carrying a load a set distance horizontally from the pelvis a flexion moment at the trunk is created (Anderson et al., 2007). Causing an equal and opposite extension moment at the trunk by the erector spinae (ES) and hip extensors at the pelvis. These moments influence the muscle activity seen in anterior load carriage, placing a higher demand on the musculature of the hip, pelvis, and trunk simultaneously in order to contribute to the increased demand for stability (Anderson et al., 2007).

Lee and Lee (2003) investigated twelve male participants while unloaded and carrying a 12 kg load above the head, at shoulder height, and at knuckle height with a wide foot placement (45 cm apart) and a narrow foot placement (20 cm apart). Each participant stood on a platform, barefoot, in an erect posture for 7 seconds followed by a functional limits of stability test. This study indicated that the wider foot placement lowered the COG_{sys} possibly contributing to increased stability.

Load carriage requires postural adaptation to maintain balance and translate the load to a new location. Research to date has focused mainly on posterior load carriage and anterior movements. However, marching band activity requires the participants to translate in all directions. In particular, snare drummers are often required to perform a "crab-walk" which is a cross-over step gait that permits the performer to move laterally while still facing the director and continuing to direct the sound forward. This project will address the unique conditions associated with marching band snare drummer load carriage including anterior load carriage, restricted arm movements and translation in multi-directions.

Section 3: Muscle Activity & Contact Pressure

The majority of load carriage literature has focused on the kinematic changes in body posture and gait dynamics as stand-alone responses. However, changes in joint and gait kinematics influence how the muscles must respond to the loading demands on the body. The following section will present the relevant literature associated with electromyography and contact pressures during load carriage.

Surface Electromyography

Posterior loads causes the COG_{person} to shift backwards creating an extension moment in the trunk (Cook & Neumann, 1987). The body responds by flexing the trunk to move the COG_{sys} closer to its unloaded position, resulting in increased levels of the trunk flexors (RA). Al-Khabbaz et al. (2008) utilized a frameless bookbag during a static task to determine the level of RA and ES activity when nineteen college-aged males stood for fifteen seconds during an unloaded and 10, 15, and 20% BM loaded condition. The researchers found an increase in only the rectus abdominus (RA) activity and no significant change in the ES activity, even though the participants in this study showed a backward inclination in posture as the load mass increased. The trunk posture inclined backwards to the level of 3.37, 3.02, and 3.90° for the 10, 15, and 20% BM load conditions respectively, resulting in increases in the right RA to levels of 35, 105, and 157% and 20, 65, and 110% for the left RA. These findings show that posteriorly loading the trunk results in trunk extension and that the most significant changes in RA activity occurred at 20% BM, supporting previous research that loads of 20% BM should be avoided. Although the ES activity levels were not significant, possibly due to the role of the ES as a trunk stabilizer and being utilized daily, the ES trends showed increasing activity as the load increased. The ES and RA were active simultaneously and showed increased levels of activity as the load increased. Circumstances in which the body will utilize the current antagonist musculature simultaneously in support of the agonist muscles of that particular task is known as coactivation (Latash, 2008).

Coactivation occurs to stiffen the joint in an attempt to compensate for the effects of a perturbation, and as a safety mechanism of the central nervous system (Latash, 2008). In the case of being externally loaded, coactivation will occur in order to help to maintain the stability of an individual. It is important to note that even though coactivation is considered a protective mechanism, the increased levels of muscle activity may contribute to the onset of muscular fatigue (Holewijn, 1990).

In a study looking at the effect of load carriage and muscle activity, Bobet and Norman (1984) investigated the muscle activity of the ES and upper trapezius (UT) of eleven males while unloaded and while carrying a fixed mass of 19.5 kg in a framed backpack. The load was placed on the posterior portion of the trunk at ear level and at the height of the xiphoid process, while each person walked 5.6 km/h along a level 90 m long course. A change in muscle activity can occur as a result of a change in the posture of the body, as well as placing the body under an external load (Bobet & Norman, 1984). The results of this study indicated that loading the body (22-34% BM of participants) at heights corresponding to ear level and the xiphoid process increased the muscle activity in both the ES and UT. Overall, the ES activity levels were lower than the UT, and between the two load placements, the lower placement (xiphoid process) produced the lowest sEMG values for each muscle group. Positioning the load at ear level resulted in the body becoming less stable compared to the unloaded condition, and this instability was compensated for by the musculature.

Motmans et al. (2006) investigated the effect of not only unframed backpacks, but shoulder bags (messenger bags), front packs, and double packs on the muscle activity levels of the two primary trunk stabilizers: RA and ES, bilaterally. Nine male and ten female college students stood erect with the feet 15 cm apart for 30 seconds, while unloaded as well as carrying a shoulder bag, backpack, front pack, double pack, all loaded with 15% BM. Similar to previous kinematic research, the double pack did not produce a significant difference from the unloaded position. Furthermore, the findings of Motmans et al. (2006), supported the work of Birrell and Haslam (2010) that double packs promote a more upright stance similar to being unloaded, due to similarities in muscle activities of the RA and ES for the double packs and unloaded conditions. During the backpack condition, the muscle activity of the ES decreased (77%) and RA increased (54%) while standing, as a result of the extension moment created by the load. When loaded with the front pack, the participant's response was to use the ES muscle group to move the COG_{sys} in a position more centered within the BOS (Bloom & Woodhull-McNeal, 1987; Bobet & Norman, 1984; Devroey et al., 2007). This resulted in activity levels of more than twice the reference (unloaded) values: 206%. The shoulder bag also showed a significant increase in ES activity, though not as high as the backpack or frontpack and also produced a significant difference between the contralateral (left) and ipsilateral (right) sides. Overall the muscles on the contralateral side had the highest activity levels (Motmans et al., 2006), as a result of the participants laterally flexing the trunk opposite from the load as noted in previous literature (Pascoe et al., 1997).

Devroey et al. (2007) also investigated the effects of wearing a frameless backpack while twelve male and eight female college students stood still for one minute and walked for five minutes loaded with 5, 10, and 15% BM. The researchers found that while standing still, increasing the load mass to 10% and 15% BM caused increased trunk flexion, pelvic anteversion, and hip flexion. While walking, increasing the load mass to 15% BM resulted in a significant increase in trunk flexion, although there were also trends towards increased pelvic anteversion and hip flexion. Devroey et al. (2007) hypothesized that the forward lean in this study occurred because the posterior location of the load caused the COG_{sys} to shift backward, resulting in the kinematic compensations of the body to move the COG_{sys} more within the BOS. These joint kinematic changes also resulted in changes in muscle activity. Activity of the RA and external obliques

(EO) increased at 15% BM compared to 0% and 5%, while the ES muscles decreased at 10% and 15% BM compared to 0% BM. The decreased ES activity demonstrated at the lowest load that the coactivation of both the abdominal and back muscles as a protective mechanism was not occurring, but instead the load was being carried "passively" according to the researchers (Devroey et al., 2007). It is also possible that the decrease in the ES activity may have occurred as a result of 1) reciprocal inhibition or 2) increasing the forward lean of the body to allow the load to be distributed over more surface area (i.e., the pelvis and trunk), thus decreasing the demand/need for ES to be activated.

Cook and Neumann (1987) also investigated the sEMG of the erector spinae during the stance phase of gait during an unloaded and loaded condition. The researchers asked twelve young-adult males and twelve females to walk 30 m, at 1.3 m/s. The load (10 and 20% BM) was placed in a wooden box, inside a framed backpack, carried contralaterally or ipsilaterally to the ES, and anteriorly and posteriorly to the chest. The ipsilateral position yielded a decrease in the muscular demand with no change between the load mass conditions. Both load conditions carried contralaterally to the electrodes on the ES resulted in a significant increase (11.69% MVIC) compared to the no-load condition, with a significant difference also noted between the load mass conditions (9.1% vs. 14.3% MVIC). Of the four load conditions, the anterior position elicited the highest sEMG activity (11.93% MVIC) from the no-load condition as well as between the load mass conditions (9.9% vs. 14% MVIC, 10% and 20% respectively). Overall, the female participants showed greater sEMG activity of the ES muscles during the loading conditions, with a 4.3% difference during the anterior load carriage position than the males (14.1% vs. 9.8% MVIC). One possible explanation for the significant increase of muscle activity during the anterior loading condition was that the anterior condition required the use of both hands. Restricting arm use was shown by Birrell and Haslam (2008) to prevent the horizontal motion of the trunk, resulting in greater excursions of the COG_{sys} and a decrease in stability. In order to compensate for the instability, the musculature must contract and is more likely to result in an earlier onset of fatigue or "fatigue-like effects" (Cook & Neumann, 1987).

Hong, Li, and Fong (2008) not only looked at load mass on muscle activity, but also load duration in children carrying bookbags up to 20% BM. The amount of time an individual wears a load is as influential as the load mass, load placement, and points of contact with the body. Children and adults do not typically wear a pack of some kind or lift a load in industry for mere seconds; therefore it becomes advantageous to investigate the effect of load duration on an individual. Fifteen male elementary-aged children were asked to wear a frameless school bookbag, unloaded (0% BM) or loaded with school supplies equaling 10, 15, and 20% BM, while walking on a level treadmill for 20 minutes at 1.1 m/s. Surface EMG signals were collected throughout the duration of the walk for the UT, RA, and lower trapezius (LT). As duration increased from zero to twenty minutes, there was a tendency of the muscle activity to increase. A significant increase was found for the LT after 20 minutes for 0% BM, 15 minutes for 15% BM, and only 5 minutes for 20% BM (fatigue levels were found at 15 minutes of the 20% BM). These results provide further support that loads weighing 20% BM should be avoided (Devroey et al., 2007; Kinoshita, 1985) and duration should not last longer than 15 minutes (Hong et al., 2008).

Harman et al. (1992) and Han et al. (1992) worked together on similar studies investigating the muscle activity of externally loaded individuals while altering either the load mass (Harman et al., 1992) or walking speed (Han et al., 1992). Both groups of researchers asked fifteen physically-fit men to walk across a level platform while carrying loads of varying mass in a military style backpack. Harman and colleagues (1992) utilized four different load conditions that equaled masses of 6, 20, 33, and 47 kg. The participants were asked to maintain a gait velocity of 1.3 m/s. As a result of altering the amount of load being carried, Harman et al. (1992) found that average sEMG of the gastrocnemius (GAS) and quadriceps (Quad) increased as the load increased, and the average sEMG of the ES doubled at the 47 kg load. These results may have occurred during the loading phase where the talocrural and tibiofemoral joints become eccentrically flexed, therefore requiring increased activity of the GAS and Quad respectively. The increased activity of the ES was expected based on the load being placed on the trunk (Harman et al., 1992), but as a trunk stabilizer, may have only responded significantly for the excessive load mass.

Han and colleagues (1992) utilized one load condition of 20 kg, but altered gait velocities to include 1.1, 1.3, and 1.5 m/s. The researchers found significant increases in muscle activity of the GAS and Quad following a 36% increase in gait velocity (1.1 to 1.5 m/s), as well as increases in the hamstrings (HAM) and tibialis anterior (TA) activity. However, there was no significant increase in the ES muscle activity regardless of the velocity, most likely due to the lower extremity musculature being primarily responsible of maintaining gait velocity (Han et al., 1992). In addition, due to there being no change in load masses between the walking speeds, the ES simply did not need to change its activity level to handle the 20 kg load. Both studies took the opportunity to collect additional gait kinematic and kinetic data, and found supporting results to the previous dynamic load carriage studies, for example: increased stride frequency, increased braking impulse, increased propulsive and medial forces, and increased % in DST.

In addition to investigating the demands of carrying a load anteriorly to the body at various heights on trunk ROM and posture, Anderson et al. (2007) investigated the activity levels of the RA, EO, biceps brachii (BB), anterior deltoid (AD), and the ES at three vertebral levels (T9, T12, and L3) during seven load carrying tasks. Eleven healthy college-aged males, were exposed to 2 load conditions (a hand-held barbell equaling 20% of participants' maximum voluntary isometric elbow flexion and a 14kg bucket) and 3 height conditions (knuckle, elbow, and shoulder height). Activity levels of the RA, EO, BB, AD and ES were recorded while holding the anterior loads, during a 6 second standing test and a 15 second walking test, at a self-selected pace on a level treadmill. Kinematic results demonstrated that while holding the barbell, trunk extension at the T9 and T12 vertebrae were greatest while standing; however, the average trunk ROM at all three landmarks was 1.5° greater while walking. In addition, carrying the barbell at shoulder height resulted in a larger trunk extension posture at the T9 and T12 vertebrae. The muscle activity during the barbell condition provided supporting evidence of the increased trunk ROM while walking. There was also an increase in muscle activity of the ES at the T9 level while carrying the barbell at shoulder height. In addition, the researchers observed that the normalized RA, EO, BB, and AD activity levels while walking were lower than all three ES levels (Anderson et al., 2007), indicating that the ES were primarily responsible for extending the trunk in order to shift the COG_{sys}

posteriorly to a more balanced location. The other musculature were occupied with holding the load (BB and AD), while the RA and EO were contributing in a supportive role (i.e., coactivation).

When standing with the 14kg bucket at shoulder height, trunk extension not only occurred for both the T9 and T12 vertebrae, but at the L3 vertebrae as well. Similar to the barbell condition, carrying the bucket at shoulder height resulted in the largest trunk extension angle while walking, and an increase in the average trunk ROM. As a result, the bucket condition also elicited a higher amount of muscle activity while walking, particularly for the ES at the T9 (35%), T12 (36%), and L3 (42%) levels, and the RA (132%). The largest RA activity occurred while standing and holding the bucket at shoulder height, and while walking and holding the bucket at elbow height.

The results of this study support the previous literature on posterior load carriage that suggests that the trunk will adopt a leaning posture when loaded in order to move the COG_{sys} closer to the center of the BOS (Harman et al., 2004; Kinoshita, 1985; Smith et al., 2006). Placing that COG_{sys} closer to the center of the BOS will therefore allow an individual to maintain a more balanced and upright posture. During load carriage situations, a decrease in stability of the trunk can occur due to the additional flexion, extension, lateral, and rotational moments that occur as a result of the load carriage system shifting the COG_{person} away from the unloaded position. The result is an increase demand placed upon the musculature that can lead to increased strain, onset of fatigue, and injuries.

Contact Pressure

Skin contact pressure can be considered one of the primary limiting factors in the ability to perform a load carriage task if the tension from the pack cannot be reduced (Holewijn, 1990). Sangeorzan (1989) discovered that applying a pressure of 75mm Hg (10kPa) on skin directly over a muscle can reduce the circulation of blood. A skin contact pressure of more than 105mm Hg (14kPa) was found to result in irritation, pale or dull skin, congestion of vessels, and if exposed for two hours, can induce subcutaneous edemas (swelling caused by fluid collection) and possible muscle fiber degeneration (Husain, 1953). Doan (1998) collected both objective and subjective data on nine military load carriage systems from various countries and found subjective responses that support the claims that pressures above 14 kPa can be a limiting factor for the user. Twenty-eight soldiers volunteered to perform a variety of activities that tested balance, mobility, and agility, at four different activity stations for two of the nine load carriage systems. At each station, the soldiers were asked to rate the load carriage system based on factors including the sense of the load being controlled, comfort, and amount of ROM of the arms and torso. The soldiers completed the four station circuit for a total distance marched of 6 km. Results showed that 95% of the soldiers reported discomfort when the average shoulder pressure exceeded 20 kPa, and 90% reported discomfort at 18 kPa. It should be noted that the load that was being carried was 24 kg, approximately 30% BM of the average mass of the participants (Doan, 1998), which is well above the recommended 10-15% BM by previous research (Malhotra and Sen Gupta, 1965; Pascoe and Pascoe, 1999; Devroey et al., 2007; Hong et al., 2008). The concern here is that the soldiers did not comment or notice any issues with the load carriage systems at the lower and still damaging levels of 10-14 kPa, indicating that some internal damage could be occurring before the individual is aware and protective mechanisms such as moving the pack to prevent blood occlusion can occur. In a study investigating the impact of load mass and carrying mode, Holewijn (1990), demonstrated that skin contact pressure (kPa) was a physical parameter that could not only limit load carrying ability but load carrying duration as well.

Husain (1953) performed a study of the legs of rats using a pressure cuff to determine the durations of pressure that could be withstood by an animal and the response it created. It was determined that low pressures (100 mm Hg vs. 800 mm Hg) maintained for long periods of time can cause more damage than high pressures for short periods of time. In addition, it was found that if the initial leg had pressure applied to it again three days after release, a much smaller pressure (50 mm Hg) applied for 1 hour caused muscular degeneration to begin to occur. Based on these findings, Husain (1953) recommended that a substantial resting period should be utilized between loading conditions. Husain (1953) also investigated 10 case studies of human patients with bed sores in order to document the duration in bed and the etiology of the musculature either beneath or near the bed sore. The researcher determined that bed sores are associated with prolonged pressure (as early as one hour), nearness to a bony prominence, or a decreased ability of the patients to withstand normal pressure thresholds as a result of multiple medical conditions. Fortunately, Husain (1953) also noticed that if it were possible to remove the excess pressure (i.e., repositioning the patient in bed) a hyperaemic effect resulted, where the blood flow was restored. Non bed-ridden individuals do this every day as second nature. For example, sitting during a two-hour lecture, a student will change position within the chair to alleviate the uncomfortable sensation in the buttocks.

The inability to relieve excess pressures has detrimental effects, the extent of which depends on the location (i.e., bed sores at the superficial level, decreased blood flow at the vessel level). If the pressure occurs at the nervous system level, the result can be decreased sensations and loss of function. Goodson (1981) determined that high skin contact pressures from tight fitting straps, such as those often used in an attempt to place the backpack load closer to the trunk, can affect the superficial nerves in the brachial plexus. This is known as Rucksack Palsy and can result in symptoms anywhere along the arm depending on the site of the pressure. If these nerves are pinched or damaged, the results can be numbness, interrupted neuromuscular signals, weakness, or loss of function. Fortunately, Goodson (1981) determined that if the pressure was removed in a timely manner, the symptoms would be alleviated and the sensations in the upper extremity would return.

Section 4: Lumbar Belt Implementation

Hiking pack research, although less abundant than bookbag studies also provides recommendations about how to handle the loads placed on the body. These include: placing the load closer to the body (Cook & Neumann, 1987) and utilizing a lumbar belt to decrease the load placed on the shoulders (LaFiandra & Harman, 2004; Mackie et at., 2005). Lumbar belts have been advertised to customers based on the subjective concern of comfort and performance. However, it is the empirical evidence that these structures are able to provide for more effective load carriage that will assist in potential adaptations to the marching snare drum system (SDS) (Bloom & Woodhull-McNeal, 1987; Holewijn, 1990; Reid et al., 2004).

Holewijn (1990) recruited four college-aged males, unaccustomed to carrying backpacks, and placed loads of 5.4 kg and 10.4 kg inside both a military backpack that placed the load on the shoulders and a custom backpack that divided the load between the shoulders and hips through the use of a lumbar belt. Each participant was asked to stand on a treadmill for 20 minutes and walk on the same treadmill at a speed of 1.3 m/s for an additional 20 minutes. Data collection consisted of an unloaded, 5.4 kg military pack, 5.4 kg custom pack, 10.4 kg military pack, and 10.4 kg custom pack. The pressure data were measured at the lateral border, medial border, and center of the right where the straps made contact with the body. Results indicate that placing a load on the body does cause an increase in skin contact pressure, particularly from the top of the shoulder along the back side of the strap towards the scapula. Across all conditions, the military pack increased skin pressure up to 203 mmHg as compared to the custom pack (only 15 mmHg) for both the static and dynamic conditions. The data also showed significant changes between the masses being carried in the military pack; for example, standing while carrying the 10.4 kg load resulted in a 36% increase in pressure compared to the 5.4 kg load. The custom pack showed significantly lower pressure on the superior aspect of the shoulder than the military pack, and the military pack produced a significant increase in pressure on the edges of the straps within loading conditions (Holewijn, 1990).

Military load carriage research not only focuses on the load mass, gait kinematics/kinetics, and temporal variables, but forces exerted on the body as well.

LaFiandra et al. (2004) chose to investigate the effect of a military backpack mass on the forces being exerted onto the individual, and how much of the total force is being exerted on the upper back versus lower back. Eleven male soldiers were recruited to walk on a treadmill for 3 minutes at 1.34 m/s while carrying a MOLLE system loaded at 13.6 kg, 27.2 kg, and 40.8 kg. Two force transducers were mounted between the connections of the lumbar belt and the pack frame to determine if the increase in force exerted on the body was due to increased load mass and to determine the amount, if any, of the vertical forces were transmitted from the shoulders to the pelvis through the use of a lumbar belt. The results indicated that increasing the load mass resulted in an increase in the vertical forces on the lower back. In particular, as the load mass increased from 13.6 (133 N), 27.2 (266 N), and 40.8 kg (400 N), the mean vertical force exerted on the lower back increased from 41.85, 74.73, and 119.71 N, while the mean vertical force on the upper back/shoulders increased from 92.36, 193.10, and 282.19 N respectively. Based on these findings, LaFiandra et al. (2004) determined that regardless of load mass condition, 30% of the vertical force generated by the pack was transmitted to the lower back with the use of lumbar belt, leaving 70% of the load to be managed at the shoulders.

Mackie and colleagues (2005) also investigated the influence of load carriage, the use of a lumbar belt, gait speed, and shoulder strap length on shoulder strap tension and shoulder pressure. Similar to Holewijn (1990), Mackie and colleagues (2005) determined that the load mass had the greatest influence on shoulder strap tension and shoulder pressure. The researchers also found that by utilizing a lumbar belt and adjusting the strap length, the load could be more effectively redistributed to the hips while the shoulders experienced less pressure.

Mackie et al. (2005) utilized a load carriage simulator with a programmable three degrees of freedom platform. Also employed was a mannequin representing a 5th percentile Canadian armed forces female to replicate the characteristics of a thirteen year old student. A commercially available backpack with shoulder straps and a waist belt was used. The load cells were located at the top and bottom of the shoulder straps to ensure that desired tension levels were achieved. Pressure sensors were also located on the top of the right shoulder, under the strap. The 2 load conditions (5.3 kg =10% BM of mannequin and 7.9 kg =15% BM of mannequin), were placed at 5.5 cm and 11 cm horizontally from the inner backpack lining, high and low on the mannequin, as well as with and without the use of a lumbar belt. Results of these simulated load carriage conditions indicated that load mass had the greatest influence on the strap forces and more importantly, shoulder pressures. However, the increase in shoulder pressures was not proportionate as the researchers had expected. Loads of 10% BM produced pressures of 222 k Pa while 15% BM produced pressures of 378 k Pa. Utilizing a lumbar belt yielded a larger contact pressure for the 10% body mass condition as compared to the non-lumbar belt condition (246 k Pa), but decreased the pressure at 15% BM to 355 k Pa. Tighter shoulder straps produced an overall pressure of 350 k Pa for the 15% BM. The looser straps resulted in less shoulder pressure (250 k Pa vs. 350 k Pa) and placing the load closer to the inside wall of the backpack also resulted in less pressure (295 k Pa vs. 305 k Pa). As a culmination of the results, when the lumbar belt and loose shoulder straps were used in conjunction, there was an overall decrease in the pressures experienced at the shoulder (Mackie et al., 2005).

Martin and Hooper (2000) also investigated the effect of relocating a portion of the load mass to the hips through the use of five backpack belt designs on shoulder pressure. The hips and pelvis can withstand greater forces (three times less sensitive to pressure) than other parts of the body due to the spherical head, deep acetabulum, and ligamentous support (Scribano, 1970). The spherical head and acetabulum allows the forces to be distributed over a larger surface area, and dissipated along the curvature of the femoral head. The use of these anatomical structures provided Martin and Hooper (2000) the basis for studying the use of a lumbar belt during load carriage.

In total, forty-one male and thirty-one female students (no documentation was provided regarding previous experience with pack carriage) volunteered to carry a 20 kg military rucksack for 30 minutes at a speed of 3.5 km/h. Shoulder and hip pressure was measured using a TekScan® F-Scan© sensor system. In addition, the participants were asked to rate the comfort of each pack/belt trial. The first belt was situated around the participants' waist and was made of foam. Belt two was also made of foam, but was placed at the hips along with belts 3-5. Belts 3, 4, and 5 were all made of plastic and various types of an air mesh material. The results indicate that belt #1 had the largest maximum pressure (139 kPa), the smallest reduction in shoulder pressure (8.7%), with the worst comfort ratings (2.85) at the shoulders. Belts 2, 3, 4, and 5 however, had maximum pressures of 82, 88, 79, and 57 kPa respectively, and were able to reduce the shoulder pressure to levels of 48.3, 42, 41.1, and 39.2% respectively (Martin and Hooper, 2000). These results provide further support that including a lumbar belt and using it appropriately has significant benefits to transferring a portion of the load to a more suitable structure such as the hips.

Reid and colleagues (2004) investigated static load carriage and the effect of pack framework by implementing lateral “stiffness” rods to a pack that already had a lumbar belt but no permanent frame. The researchers investigated the forces being exerted on the body at the shoulders and waist to determine if the application of “stiffness” rods could redistribute the load of the pack vertically, in fact shifting the load lower. Reid et al. (2004) hypothesized that transferring the forces to the pelvis would minimize the horizontal forces in the lumbar region and would stabilize the load. A mannequin was utilized and held at 10° of trunk flexion. Reid et al. (2004) found that by attaching a rod from the pack to the lumbar belt, allowed the rod to become an “active component of system”, transferring forces from the shoulder straps to the lumbar belt. The rods reduced the vertical force applied to the upper body by 14% without a difference in shear forces on lumbar spine (L3/L4). The rods also decreased the tension on the shoulders, and decreased the flexion moment at the lumbar (Reid et at., 2004), both adaptations that would presumably decrease the muscular demand at the shoulders and trunk respectively (Goh et al., 1998).

In summary, the inclusion of a lumbar belt decreases the forces imposed at the shoulder. Therefore, it is reasonable to conclude that the inclusion of a lumbar belt to the marching carrier will decrease the forces noted at the shoulder. It is less clear as to the influence of the lumbar belt on gait mechanics as the pelvis will already be constrained by the location of the drum.

Section 5: Summary

Previous research has shown that load carriage influences COG_{sys} excursion lengths, sagittal trunk movement, hip flexion, and knee flexion angle (Anderson et al., 2007; Kinoshita, 1985; Schiffman et al., 2007). One outcome that occurs as a result of an external load includes shifting the COG_{person} closer to the edges of the BOS (increasing sway) and thus compromising the balance of the individual. In response to an external load being placed on the front or back of the body, individuals in a non-restricted task can utilize the lower extremities for shock absorption, and lowering the COG_{sys} for increased stability and balance (Devroey et al., 2007; Kinoshita, 1985; Ling et al., 2004). However as noted in Table 1, the marching band drummer is restricted when loaded and can't always take advantage of these adaptations.

These limitations emphasize the gap in the literature and the inability to automatically apply the solutions found in the load carriage literature to marching load carriage tasks. The first is based on the recommendation that safe load carriage should be between 10 and 20% BM (Hong et al., 2008; Moore et al., 2007). However, because the mass of the SDS remains constant (8.58kg) any adjustment to this load is unreasonable. A second constraint is that the horizontal placement of the COG_{drum} cannot be adjusted, and the rods on the carrier are not load-bearing rods similar to those found in framed packs. The inability to move the COG_{drum} closer to the individual limits the opportunity to decrease the horizontal shear forces from the marching carrier at the lumbar region. Without load-bearing rods in the carrier, the shear forces continue to act on the lumbar region instead of being transferred to the rods. A third restriction is the length of the body of the carrier. The length is standard but drummers are many different heights. The hiking and backpack industry suggest that the size of the pack should fit the torso (REI, Inc., 2010). However, all drums must be maintained at the same level for aesthetic reasons. This often precludes vertical adjustments of the load on shorter individuals as moving the drum too high would cause the drummers' arms to be placed in an awkward and unrealistic playing position. Arm placement/location is another constraint that has an impact of the performance ability of marching drumline members. Research that investigated rifle carriage and the restrictive arm motion that accompanies that load (Attwells et al., 2006; Birrell & Haslam, 2009), relates well to the restricted arm motion found in marching band participants. This limited arm motion has been shown to increase the movement of the COG_{person}, leading to increases in the side-to-side motion which could lead to a loss of balance and potentially a fall (Birrell, Hooper, & Haslam, 2007). Additional kinematic changes that individuals wearing the carrier cannot employ are included in the following table (**Table 1**) including the possible implications and outcomes as a result of wearing an SDS.

Table 1
Kinematic Challenges of the SDS

Responses During LC	Limitations of Drummer	Implications/Outcomes
Increased Hip and Knee Flexion[2,3,5,7,8,9,11,15]	Limited due to location of the drum	*Limited ROM[4] *Increased muscle activity[1,2,6,7,9]
Increased Pelvic Anteversion[4,7]	Limited due to the location of the drum	*Limited ROM[4] *Increased muscle activity[1,2,6,7,9]
Trunk Lean to Shift the COG[1,2,3,7,15]	Limited due to aesthetic requirement of posture	*Increased muscle activity[1,2,6,7,9]
Natural Arm Movement[4]	Limited due to requirements of playing	*Increased muscle activity[1,2,6,9] *Inability to assist in modulating COG excursion[11,19]
Decreased SL[4,13,18]	Limited due to standardized step sizes	*Limited ROM[4]
Increased SF[3,13,18]	Dictated by the music	*Decreased time spent in DST
Increased DST[4,8,12,15]	Dictated by the music	*Decreased stability[8]
Implementing a Hip Belt[10,14,16,17,19]	Limited due current design	*Increased muscle activity[1,2,6,7,9,19] *Increased contact pressure[10,16,17] *Decreased stability, increased excess motion[19]

[1]Al-Khabbaz et al., 2008; [2]Anderson et al., 2007; [3]Attwells et al., 2006; [4]Birrell & Haslam, 2009; [5]Cham & Redfern, 2004; [6]Cook & Neumann, 1987; [7]Devroey et al., 2007; [8]Goh et al., 1998; [9]Harman et al., 1992; [10]Holewijn, 1990; [11]Holt et al., 2003; [12]Kinoshita, 1985; [13]LaFiandra et al., 2003; [14]LaFiandra et al., 2004; [15]Ling et al., 2004; [16]Mackie et al., 2005; [17]Martin & Hooper, 2000; [18]Martin & Nelson, 1986; [19]Sharpe et al., 2008

There is much to be understood regarding the load carriage demands and constraints placed upon the marching snare drummer. Including but not limited to: (a) determine the adaptations (postural and kinematic) in experienced marching band snare drummers to the load carriage and constraints of wearing a marching snare drum system: and (b) determine the influence of a specialized lumbar belt on these variables. Backpacks that utilize lumbar belts have been shown to redistribute as much as 30% of the vertical force from the upper back to the lower back for better load distribution and possibly less muscular demand on the participant (LaFiandra & Harman, 2004). These findings suggest that the inclusion of a frame and lumbar belt can result in some of the most positive combined effects to backpack load carriage, and perhaps provide guidance for approaching marching load carrying tasks. The outcome of this research will advance the load carriage literature and provide potential solutions to the unique population of marching band snare drummers.

CHAPTER III

METHODS

Thousands of marching band musicians perform demanding and strenuous movements each day, and a particular group of musicians perform these movements while carrying large cumbersome loads. These load carriage demands were investigated in order to determine the influence of a SDS on the biomechanics of the marcher and to provide information about a population and load carriage technique that has received little attention in the literature. Therefore, the purpose of this investigation was three-fold: 1) to determine the influence of a SDS on trunk and lower extremity kinematics while standing and marching, 2) to determine the contact pressures and muscular demand placed on the erector spinae individuals wearing a SDS, and 3) to investigate the influence of a lumbar belt (modified SDS (MDS)) on trunk kinematics, muscular activity, and contact pressures. The following chapter presents the methodology used to address the purposes of the present study and includes the following sections: (a) participants, (b) equipment, (c) procedure, and (d) statistical analysis. The research protocol was approved by the Auburn University Institutional Review Board for Research Involving Human Subjects.

Participants

Sixteen college-aged males (N = 12) and females (N = 4) between the ages of 19-25 years, with drumline marching experience, specifically snare or tenor drum experience, were recruited from the 2010-2011 Auburn University Marching Band. Although there are only 9 individuals who currently participate in the marching band as snare drummers, other drumline members were recruited as long as they had experience marching at least one season (college or high school) as a snare or tenor drummer. The tenor drums also place the load anteriorly and inferiorly to the drummer's body and therefore have a similar load demand on the body. All participants were clothed in a compression tank-top, compression shorts, and tennis shoes. Each participant completed a health screening survey (**Appendix** A) and was excluded if a question was answered "yes" and it was deemed by the primary investigator that the "yes" would influence the participant's performance in the study. Exclusion criteria included any current lower extremity, trunk, or shoulder injury or surgery, any known decrements in balance, or currently had an inner ear disturbance. Each participant was asked to sign a University approved informed consent document to indicate his or her voluntary willingness to participate in this project (**Appendix B**).

Setting

All testing and data collection occurred in the Sport Biomechanics Laboratory at Auburn University (1127 Memorial Coliseum). The Sport Biomechanics Laboratory is a large enclosed laboratory with the necessary equipment to carry out this research.

Materials

Marching Snare Drum System

The SDS that was used for this study and by the Auburn University Marching Band Drumline is comprised of the Pearl® (Pearl® Corporation, Nashville, TN, USA) Championship Maple FFX-1412 Snare Drum (7.67 kg) (**Figure 1**) with a Randall May® (Randall May® International, Inc., Irvine, CA, USA) Aluminum Tube Snare Carrier (0.91 kg) (**Figure 2**). The maximum length of the carrier from the peak curvature of the shoulder frame to the lowest bracket arm position is 78.1 cm (**Figure 3**), the height of the snare drum is 35.6 cm (**Figure 3**). When worn together the total length of the SDS from the top of the shoulder frame to the bottom of the drum is approximately 101 cm. The reason why the total length does not equal 113.7 cm is due to the 12.7 cm where the top of the drum sits above the bracket arm (**Figure 4**). The shoulder frame is 40.64 cm in length, 5.1 cm in width, with a back piece that makes contact over the scapulae that is 25.4 cm in length (**Figures 5 & 6**). The abdominal plate was measured at its three widest parts: top (13 cm), middle (24 cm), and bottom (29 cm) (**Figure 7**).

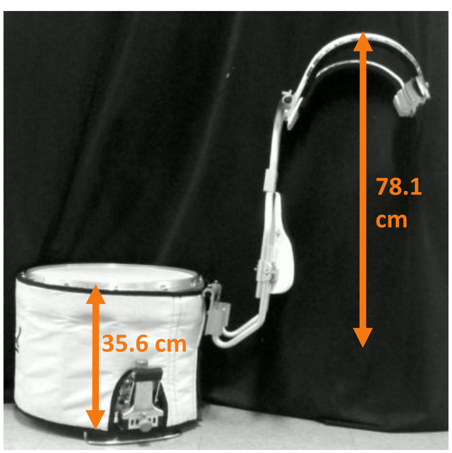


Figure 3. Length of the carrier and height of the drum.

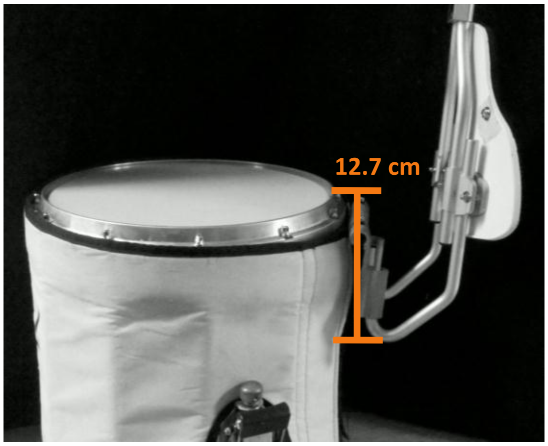
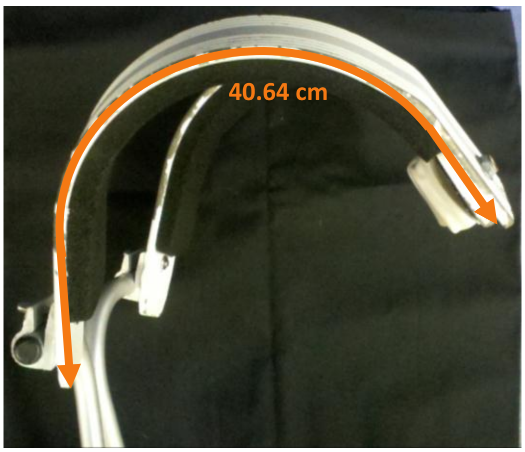
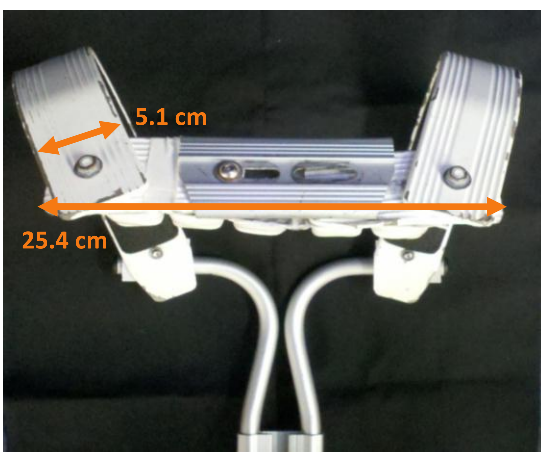


Figure 4. Distance between the bottom of the bracket arm and top of the drum.





Figures 5 & 6. Snare drum carrier shoulder frame from the sagittal and posterior view respectively.

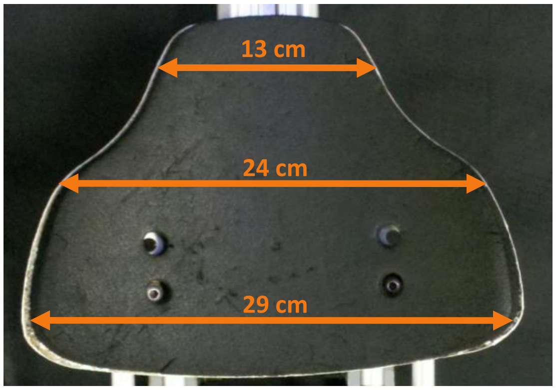


Figure 7. Snare drum carrier abdominal plate from a posterior view.

Each participant used the same snare drum set, and the drum height was restricted to a common height based on recommendations by the percussion instructor, as would be the case during the regular marching season.

Lumbar belt

The belt for this study will be referred to as lumbar belt as that is the location on the body upon which it will be placed. The lumbar belt was constructed by the primary investigator using 2 mm neoprene (Seattle Fabrics, Inc.®, Seattle, WA, USA), 0.95 cm polyethylene tubing (Watts®, N. Andover, MA, USA), one roll of 1" x 60" nylon strapping (Sew-Ology®, Oklahoma City OK, USA), ½" closed cell foam (Seattle Fabrics, Inc.®, Seattle, WA, USA), and industrial strength velcro (10.1 cm x 5 cm) (Velcro USA Inc.®, Manchester, NH, USA).

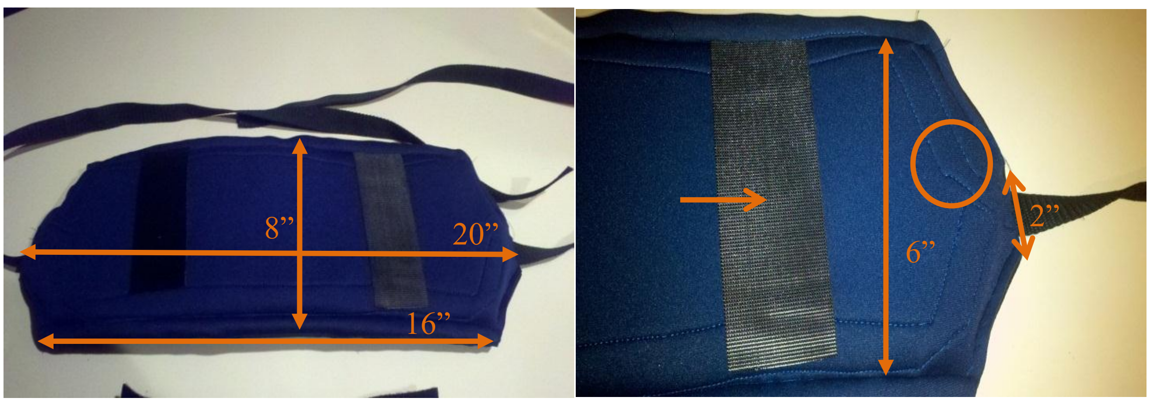
The initial construction consisted of the two neoprene layers were cut into a standard lumbar belt shape and consisted of these specifications:

- Maximum length of the neoprene is 20" and maximum height is 8" (**Figure 8a**)
- Minimum height of the neoprene is 2" tall located at the very ends of the material (**Figure 8b**)

- The neoprene maintained the height of 8” for the length of 16” (8” left and right of the center and in line with the plastic tubing), at which time was cut at a 60° angle (**Figure 9**)
- Three pieces of neoprene material used for this belt; one solid piece and two smaller pieces that overlapped each other. The largest piece had the dimensions described above and was the layer that faced away from the individual. (**Figures 8a & 8b**) The second and third pieces were also cut to a length of 20” but only to a height of 5” and 4” respectively. This one inch difference allowed for the overlap between the two pieces for strength, protection of the foam, and easy removal of the foam in order to allow the belt to be washed. (**Figures 10a & 10b**)
- The foam was cut to a length of 18” and 6” tall, following the path and angle of cut of the neoprene material, leaving 1” on either end for the space between the foam and the neoprene where the nylon strapping would exit the belt. (**Figure 10a**)
- The plastic tubing was cut to a length of 16” and placed at the top and bottom edges of the material just above the foam. A pocket was then sewn at the top and bottom edges of the belt to encase the plastic tubing and allow for easy removal. (**Figure 11**)
- The nylon strapping was cut into two-30” pieces. One piece was sewn at one end of the belt, angled up towards the outer edge, sewn in parallel to the plastic tubing, angled again towards the opposite end where it exited to be used as one of the ends that would go around the metal tubing of the drum carrier. This process

was repeated with the other 30” piece with the only different being the opposite starting position. (**Figures 8a, 8b, & 11**)

- Velcro was attached in three places for this belt-1) along the straps that exited the belt and were used to wrap around the drum carrier (**Figure 12**); 2), along the back of the belt to provide for attachment sites for the straps (**Figures 8b, 11, & 12**); and 3) along the side facing the individual, on both sides of the neoprene pieces that overlapped each other for insertion and removal of the foam. (**Figures 10a &10b**)



Figures 8a & 8b. Lumbar belt construction. Figures 8a and 8b note the overall size of the belt, as well as the particular stitching of the nylon straps (circle) and the placement of the outside velcro (single arrow).

Figure 9. Figure 9 notes the angle at which the corners of the belt were cut at.

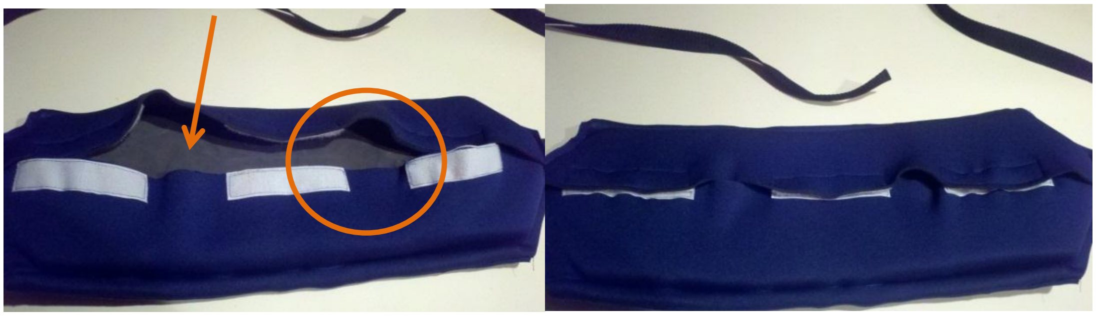


Figure 10a & 10b. Lumbar belt construction. Figure 10a notes the two separate layers of neoprene (circle) that were folded over each other to provide a secure closure (10b), as well as easy opening to insert and remove the foam (arrow). The foam was cut to a length of 18”, a max height of 6”, min height of 2” at the ends, and a 60° angle taper from the center to the ends.

Figure 11. Lumbar belt construction. Figure 11 notes the plastic tubing used for the belt design, and the pocket that was stitched for its insertion and removal. Seen here again are the stitching techniques for securing the nylon strapping and the velcro attachment.

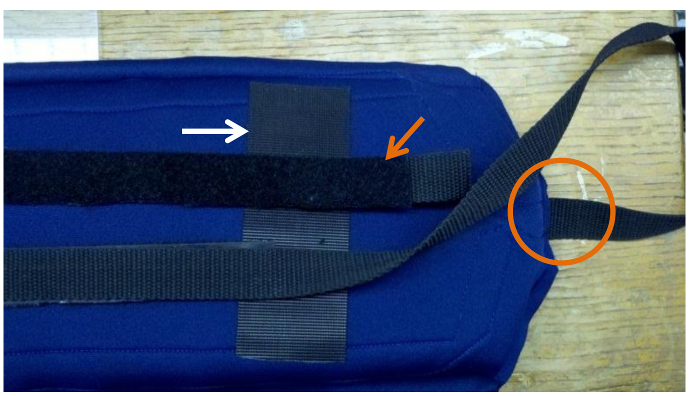


Figure 12. Lumbar belt construction. Figure 12 notes the velcro attachment to the straps (orange arrow) seen exiting the belt (circle), as well as the back of the belt attachment (white arrow) to which the straps will connect to for a secure hold of the drum to the individual.

Instrumentation

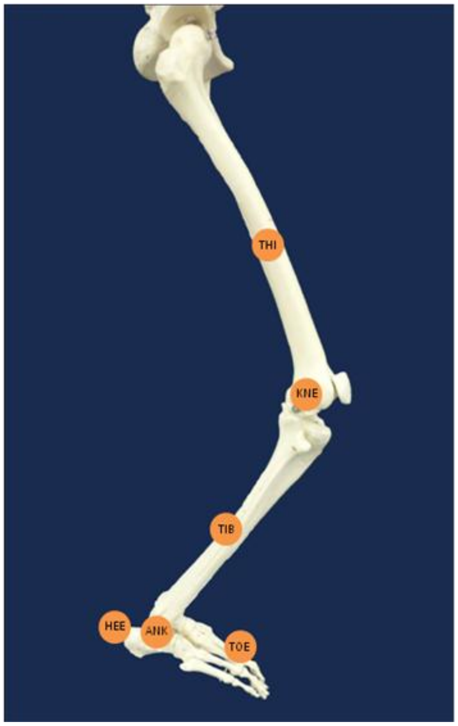
Kinematics

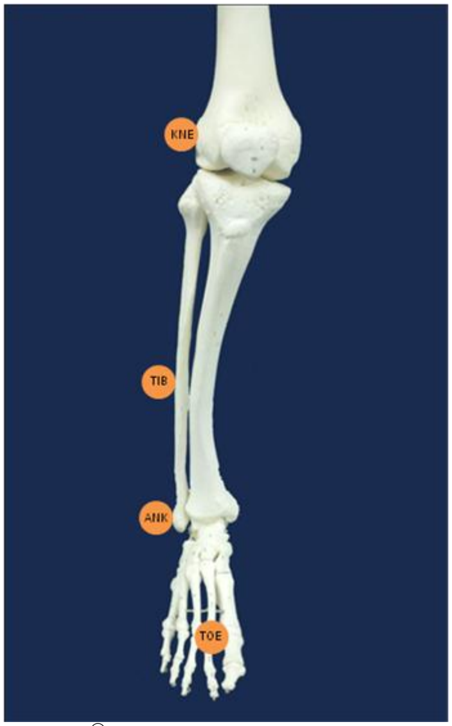
Three dimensional trajectories of the foot, ankle, knee, hip, and trunk were obtained with a ten camera Vicon® MX motion analysis system (Vicon®, Los Angeles, CA, USA) with a sampling frequency of 288 Hz (Devroey et al., 2007). To track the movement strategies of the lower extremities and trunk, 34 retroreflective markers were attached to each participant's body (**Table 2 and Figures 13-16**). The markers used were spherical 14mm retro-reflective markers (MKR-6.4©, B&L Engineering®, Tustin CA, USA) attached with double-sided tape (CLEAR-1R36©, Hair Direct Inc.®, Bainbridge, PA, USA). The standard Plug-In-Gait© marker placement protocol (Vicon®) was used with the sacrum marker being used in place of a left and right PSIS in order to accommodate the drum carrier and lumbar belt placements.

Table 2
Names and positions of markers used in Plug-In-Gait© Model

Marker Name	Position	Segment
R/LFHD	Right/Left Front of Head	Head
R/LBHD	Right/Left Back of Head	Head
CLAV	Sternoclavicular Joint	Chest
STRN	Sternum	Chest
C7	7th Cervical Vertebrae	Spine
T10	10th Thoracic Vertebrae	Spine
RBAK	Right Midtorso	Asymmetry Identification
R/LSHO	Right/Left Acromioclavicular Joint	Shoulder
R/LELB	Right/Left Lateral Aspect of Humeroulnar Joint	Elbow
R/LWRA	Right/Left Radius	Wrist
R/LWRB	Right/Left Distal End of Ulna	Wrist
R/LFIN	Right/Left 3rd Metacarpophalangeal Joint	Knuckle/Finger
R/LASI	Right/Left Anterior-Superior Iliac Spine	Pelvis
SACR	Sacrum	Pelvis
R/LTHI	Right/Left Lateral Aspect of Femur	Upper Leg
R/LKNE	Right/Left Tibiofemoral Joint	Knee
R/LTIB	Right/Left Lateral Aspect of Fibula	Lower Leg
R/LANK	Right/Left Lateral Malleolus	Ankle
R/LTOE	Right/Left Distal End of 2nd Metatarsal	Toe/Foot
R/LHEE	Right/Left Calcaneus	Heel

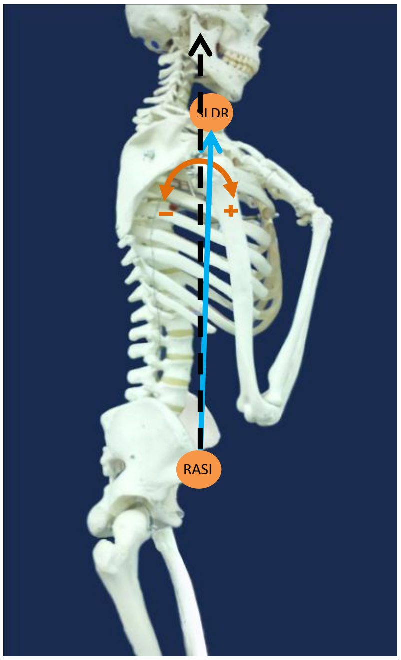
Figures 13 & 14. Anterior and posterior views of the Plug-In-Gait marker locations.

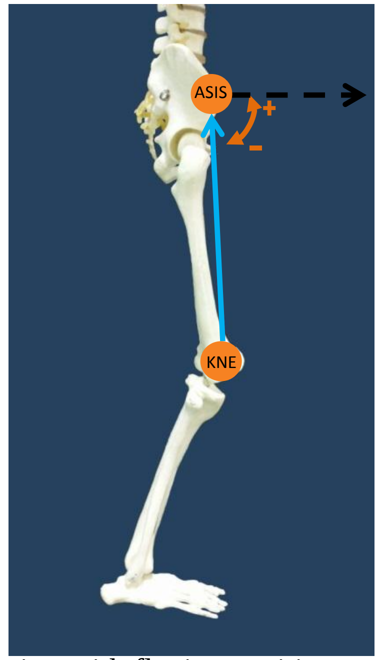




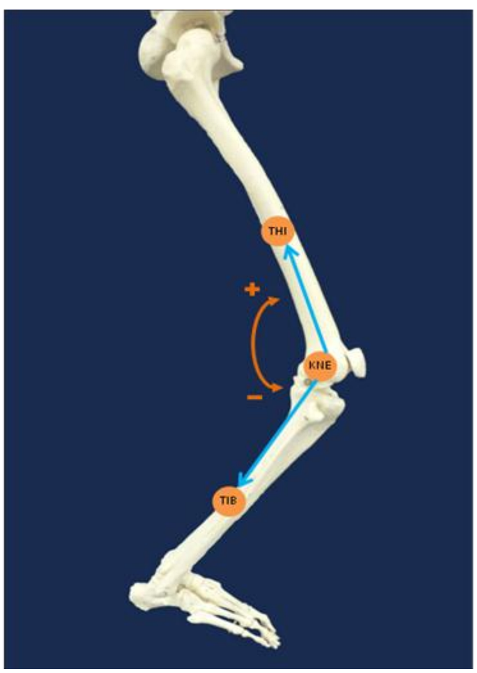
Figures 15 & 16. Sagittal views of the Plug-In-Gait© marker locations.

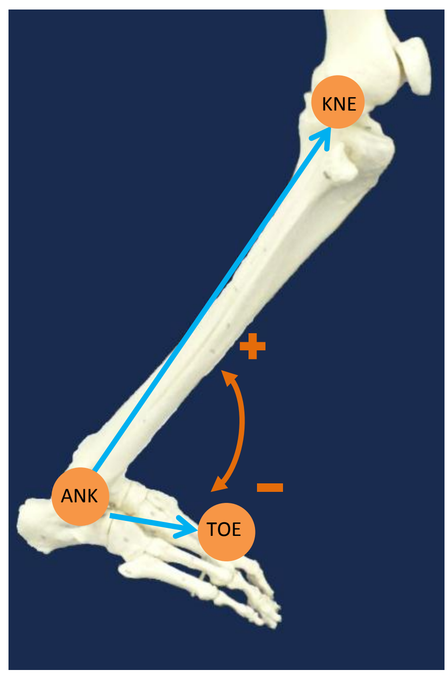
Joint angles were calculated by Vicon® software using a modified Plug-In-Gait model (Devroey et al., 2007) (**Figures 17-20**).





Figures 17 & 18. Trunk and hip flexion/extension with flexion positive and extension negative relative to the absolute angle (Attwells et al., 2006).





Figures 19 & 20. Knee flexion/extension with flexion positive and extension negative. Ankle plantarflexion and dorsiflexion with dorsiflexion positive and plantarflexion negative (Attwells et al., 2006).

Ten cameras were positioned to record all three cardinal planes (frontal, sagittal, and transverse) of the participant during one full stride, defined as right heel contact to right heel contact (**Figure 21**).

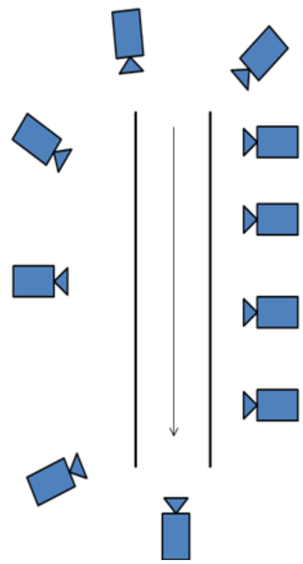


Figure 21. Vicon motion capture camera setup.

The gait cycle used for the current study was defined as right foot contact, right foot midstance, right foot heel-off, left foot contact, right foot toe-off, left foot midstance, left foot heel-off, and a second right foot contact (J. Perry, 1992) (**Figure 22**). This cycle was used because it incorporates both phases of a gait cycle (stance and swing). Both the right and the left leg are needed and consequently included for one complete stride, and therefore, periods during the gait cycle can be evaluated. The participants were asked to walk utilizing a specific marching gait (glide step) in the laboratory instead of on the marching field or road (parade comparison) at a set frequency (144 beats/min) that most resembles performance tempos.

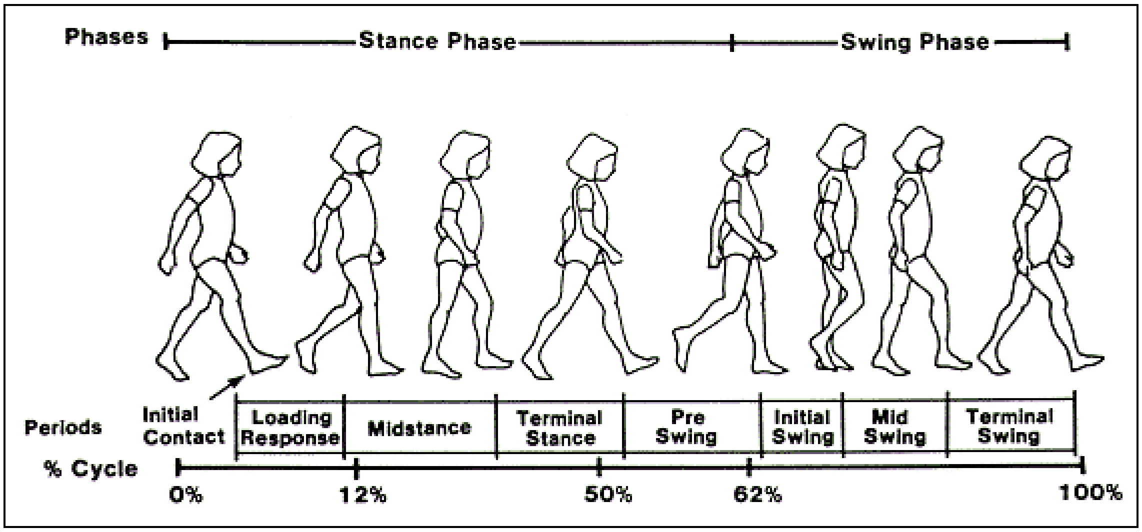


Figure 22. Functional phases of gait (Kirtley, 2008)

Surface Electromyography (sEMG)

Muscle activity was collected using **1**) Noraxon® MyoSystem 1200 tethered surface EMG receiver (Noraxon® U.S.A. Inc., Scottsdale, AZ, USA) (**Figure 23**), and **2**) .Noraxon® Telemyo 2400R-World Wide Telemetry receiver (Noraxon® U.S.A. Inc., Scottsdale, AZ, USA) (**Figure 24a**), connected via preamplifier leads to a Noraxon® Telemyo 2400T V2 wireless transmitter (Noraxon® U.S.A. Inc., Scottsdale, AZ, USA) (**Figure 24b**). The electrodes were Trace-Rite® Ag/AgCl ECG monitoring surface electrodes (Trace-Rite®, Bio-Detek Inc., Pawtucket, RI, USA) (**Figure 25**). The software utilized for data collection and processing was the MyoResearch XP Master Edition© 1.08.15 (Noraxon® U.S.A. Inc., Scottsdale, AZ, USA). The EMG signal was sampled at 1000Hz, and post processing included full wave rectification and a finite impulse response (FIR) filter (gain set at 1000, bandpass of 10-350Hz, and a notch filer of 59.5-60.5Hz). For the standing condition, the mean linear envelope was calculated for 1s. For the gait condition, the mean linear envelope of the right muscles were obtained for one stride length (Devroey et al., 2007).

Figures 23. Noraxon MyoSystem® 1200 Tethered Surface EMG receiver

Figure 24a & 24b. Noraxon Telemyo® 2400R –World Wide Telemetry Receiver & Noraxon® Telemyo 2400T V2 wireless transmitter

Figure 25. Erector spinae sEMG electrode placement

Drum Carrier Pressure

Pressure was measured using a Tekscan® I-Scan System® (v 6.10.1) pressure distribution measurement system (Tekscan®, Inc., South Boston, MA, USA) (**Figure 26**). This Tekscan® system consists of three separate pressure sensors, allowing the researcher to collect data simultaneously from the superior aspect of both shoulders (Sensor #9801) and well as the lower abdominal/pelvic girdle area (Sensor #9830) (**Figures 27 & 28**). The software utilized for data collection was the Tekscan® I-Scan® Research version 6.10.1 with Sensor Map #9801 and #9830 enabled (Tekscan®, Inc., South Boston, MA, USA). Both average and peak contact pressures were measured in the areas between the carrier and the participant, specifically both shoulders, and the abdominal/pelvic region. Average pressure was calculated by the total force exerted on the sensor divided by the number of individual cells that are being triggered in a specific body region. Peak pressure refers to the highest pressure recorded for a given pressure sensel (Bossi et al., 2000). Before each trial, the sensors were calibrated to the load (8.58 kg) of the drum system, and placed on the inside of the carrier's shoulder regions (**Figure 5**) and abdominal plate (**Figure 7**). The placement of the sensors (row and column designation) were marked with chalk onto the compression tank-top (Martin and Hooper, 2000). The I-Scan® Research v. 6.10.1 system sampled/scanned at 160 Hz.

Figure 26. Tekscan® I-Scan System® (v 6.10.1)

Figure 27. Tekscan® I-Scan System® (v 6.10.1) Pressure Sensor #9801

Figure 28. Tekscan® I-Scan System® (v 6.10.1) Pressure Sensor #9830

The sEMG and joint/gait kinematics were synchronized with a Vicon® MX Control 64-channel A/D board (Vicon®, Los Angeles, CA, USA) and all trials were videotaped with a Canon® 3CCD Digital Video Camcorder GL2 NTSC (Canon® U.S.A., Inc., Lake Success, NY, USA) for motion verification. Kinematics were analyzed with the Nexus Polygon Software© (Vicon®, Los Angeles, CA, USA). All data were saved on a computer that is stored and locked inside the Sport Biomechanics Laboratory for post processing.

Design and Procedures

The participants reported to the Sport Biomechanics Laboratory for one day of testing. After arrival to the lab, each participant was given the Auburn University Institutional Review Board approved informed consent document and preliminary medical questionnaire (**Appendix A**). By signing the informed consent, participants indicated voluntary participation in the project. The medical questionnaire was used as a screening device to eliminate anyone who had 1) sustained a lower extremity, trunk, or shoulder injury within the last six months, 2) an inner ear disorder or balance decrements, or 3) an allergy to adhesives.

After meeting the criteria for participation, both the mass and height of the individuals were measured for normalizing purposes. The participants were then asked to change into spandex shorts and tank-tops to minimize clothing artifact and disrupting the pressure sensor readings. Anthropometric information was collected and entered in the Vicon® software.

The skin was prepared for surface electromyography electrode placement following the recommendations of the Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM) (Hermens et al., 2000). The skin area was abraded to reduce the electrical impedance of the skin and cleaned with a 70% alcohol solution. The electrodes were placed at a 2cm inter-electrode spacing, 3cm lateral to the midline (Anderson et al., 2007; Seroussi & Pope, 1987). The erector spinae (ES) at the L1 level was identified by palpation and manual muscle testing as the participant performed trunk extension from a supine position (**Figure 25**). Electrodes were placed on the muscle belly parallel to the direction of the muscle fibers. A ground electrode was placed on the bony aspect of the C7 vertebrae. Maximum voluntary isometric contraction-used for normalization (MVIC) was then collected.

Thirty four trial markers were affixed to the spandex clothing and to the skin with double-sided tape and placed according to the requirements of the Vicon® Plug-In-Gait© model with slight modifications to accommodate the drum carrier and lumbar belt. A static capture was performed while the participants stood in the center of the capture volume in an anatomical T-pose position with shoulders abducted and hips at shoulder width apart. The markers were identified and labeled to comply with the Vicon® Plug-In-Gait© model modified for this study.

The pressure sensors were taped underneath the shoulder carrier and abdominal plate of the drum carrier at the beginning of each participant's data collection session. Two pressure sensors (#9801) were placed directly over the location where the SDS/MDS made contact with the body: over the superior aspect of the shoulders, angled towards the midline, crossing over the clavicle lateral from the sternum (Martin and Hooper, 2000) (**Figures 29 & 30**). The third sensor (#9830) was placed across the lower abdominal/pelvic region, directly over the location where the SDS/MDS makes contact with the body. The midline of the sensor was placed at the midline of the body (**Figures 29 & 30**).


Figures 29-30. Tekscan I-Scan System® Pressure Sensors #9801 (shoulders) and #9830 (abdominal/pelvic area) with reference to the carrier placement respectively.

Procedures

These procedures were consistent throughout all three loading conditions (unloaded, SDS, MDS), beginning with the unloaded (U) condition followed by a random assignment to the SDS or MDS loading conditions. The only difference that should be noted is the inclusion of contact pressure measurements during the SDS and MDS conditions that were taken before and after the stationary trials.

1) Participants stood for one minute while sEMG and joint kinematics were collected during last five seconds. While standing, the participants played a warm-up exercise on the drum that is a standard amongst drumlines called "8 on a hand".

2) Participants then marched through the capture volume (glide step gait) while sEMG, and joint/gait kinematic data were collected. The participants continued to play the warm-up exercise on the drum throughout the data collection. Participants were asked to march a 57.15cm step size (standard "8-to-5" marching step), and the step frequency was set at 144 beats/minute.

3) Participants then performed a marching task for one minute in the gymnasium located adjacent to the Sport Biomechanics Laboratory.
    a. The participants marched in a straight line incorporating the glide step gait and a mark time (marching in place) phase, while playing the warm-up exercise on the drum. Participants were asked to march a 57.15cm step size (standard "8-to-5" marching step), and the step frequency was set at 144 beats/minute.
4) Following the marching task, the participants reentered the laboratory and marched through the capture volume (glide step gait) while sEMG and joint/gait kinematic data were collected. The participants continued to play the warm-up exercise on the drum throughout the data collection. Participants were asked to march a 57.15cm step size (standard "8-to-5" marching step), and the step frequency was set at 144 beats/minute.
5) Participants were then asked stand in a stationary position again for one minute while surface EMG and joint kinematics were captured during the last five seconds.

Pressure data, indicated above, was collected before and after the last stationary trial during the SDS and MDS conditions. Contact pressure procedures consisted of the participants standing for one minute while playing the standard "8 on a hand" warm-up exercise. Data were collected during the first 5 seconds after being loaded and at 55 seconds before the first stationary trial, and again at 55 seconds after the last stationary trial.

During the stationary trials, the participants played the same warm-up exercise on the drum as during the dynamic trial, with the feet externally rotated to face a 45° angle. During the unloaded condition, the arms were in the playing position to control for arm placement throughout the conditions, as well as to mimic the requirements that occur during practice of "air playing" when learning new music. There was a period of at least 5 minutes of rest in between each loading condition.

Experimental Design & Data Analysis

All statistical analyses were conducted using SPSS software (version 17.0; SPSS Inc., Chicago, IL, USA) and an alpha level of statistical significance was set *apriori* at $p \leq 0.05$ ("SPSS for Windows, 2007"); all data were stored in an Excel database and imported into SPSS for analysis. During stationary trials, the data for sEMG and joint kinematics was analyzed for one second during the last five seconds of a minute trial. For contact pressure, the data were analyzed during the initial five seconds of a one minute test (Pre5), the last five seconds of a minute test (Pre55), and the last five seconds of a minute test after the marching exercise (Post55).

To investigate the effects of load carriage on EMG_{AVG} and then EMG_{PEAK}, the data were analyzed by conducting a 2 (movement) x 3 (loading condition) x 2 (time) repeated measures ANOVA. The independent variable of movement had two levels: stationary and dynamic. The independent variable of load had three levels: unloaded (body mass-BM), standard snare drum system (SDS), and modified snare drum system (MDS). The independent variable of time had two levels: pre and post.

To investigate the effects of load carriage on contact pressure (kPa) at the shoulders and then the abdominal/pelvic region, the data were analyzed by conducting a 2 (loading condition) x 2 (body placement) x 3 (time) repeated measures ANOVA. The independent variable of load had two levels: SDS and MDS. The independent variable of body placement had two levels: shoulders and abdominal/pelvic region. The independent variable of time had three levels: Pre5, Pre55, and Post55.

To investigate the effects of load carriage on joint kinematics, the data were analyzed by conducting a 3 (loading condition) x 2 (movement) x 2 (time) repeated measures MANOVA. The independent variable of load had three levels: unloaded (body mass-U), standard snare drum system (SDS), and modified snare drum system (MDS). The independent variable of movement had two levels: stationary and dynamic. The independent variable of time had two levels: pre and post. The dependent variables were trunk angle (TA_{AVG}), hip angle (HA_{AVG}), knee angle (KA_{AVG}), and ankle angle (AA_{AVG}).

Knee angle (KA_{AVG}) and ankle angle (AA_{AVG}) were also investigated during the foot flat phase of the gait cycle. Two 3 (loading condition) x 2 (time) repeated measures ANOVAs were conducted with the same independent and dependent variables discussed above.

For all statistical analyses, post hoc repeated measures ANOVAs were performed for each dependent variable of the multivariate repeated measures ANOVA that demonstrated statistical significance. Follow-up pairwise comparisons were then performed for significant findings from the ANOVA analyses to determine statistical significance between the levels of the independent variables.

CHAPTER IV

RESULTS

This study was designed to determine the influence of marching band snare-drum carriers on marching percussionists, who often complain of shoulder and low back discomfort. Specifically, the purpose of this investigation was addressed in three questions: 1) to determine the influence of a standard marching snare drum system (SDS) and a modified drum system (MDS) on the trunk and lower extremity kinematics while standing and playing (stationary) and marching forward and playing (dynamic), 2) to determine the muscular demands on the erector spinae (ES) of individuals wearing an SDS and a MDS, and 3) to determine the contact pressures at the shoulders and abdominal region of individuals wearing an SDS and a MDS. The following chapter presents the results of the current study and includes the following sections: Section 1: Participant demographics, Section 2: The effect of the SDS and MDS systems on joint kinematics, Section 3: The effect of the SDS and MDS systems on ES muscle activity, and Section 4: Difference in contact pressures between the SDS and MDS loading conditions, at the shoulder and abdominal regions.

Section 1: Participant Demographics

Sixteen members (12 males, 4 females) of the Auburn University Marching Band Drumline volunteered for participation in the current study and met the initial qualifications. Upon answering the medical questionnaire, all volunteers were included in the study and each participant signed an Informed Consent document approved by the University Institutional Review Board for Human Subjects (**Appendix B**). The averages of the participants' demographics are summarized in Table 3.

Table 3
Participant Demographics

	Mean	Standard Deviation
Age (years)	20.81	1.22
Height (m)	1.79	0.08
Mass (kg)	81.64	23.07
Years Marched	6	2.28
Years Marched Snare/Tenor	3.63	2.13
Drum Mass to BM (%)	11.11	2.35

Section 2: Joint Kinematics

In an attempt to understand the demands placed upon the human body when one wears a SDS, the angular kinematics of the trunk, hip, knee and ankle were investigated. These data were then compared to an unloaded condition and a loading configuration that included a lumbar belt added to the SDS, which created the MDS condition. Next, the angular kinematics were investigated during a stationary condition as well as during a marching forward condition.

In order to determine significance, a 3 (condition: unloaded (U), SDS, MDS) x 2 (movement: stationary, dynamic) x 2 (time: pre, post) multivariate repeated measures (MANOVA) across all four joint angles (Trunk Angle-TA_{AVG}, Hip Angle-HA_{AVG}, Knee

Ankle-KA_{AVG}, & Ankle Angle-AA_{AVG}) with a p value set *apriori* at $p < 0.05$ was employed and yielded a significant interaction effect for loading condition (U, SDS, and MDS) and movement (stationary versus dynamic) on the kinematics of trunk angle (TA_{AVG}), hip angle (HA_{AVG}), and ankle angle (AA_{AVG}), (Wilks's $\Lambda = 0.100$, $F(2,14) = 10.297$, $p < 0.001$, $\eta^2 = 0.684$, *Power* = 1.000). This indicates that different loading conditions altered joint kinematics differently across movement conditions. Descriptive statistics for all dependent variables are presented in **Appendix C**.

Average Trunk Angle Over One Stride

The MANOVA revealed that there was a significant interaction between loading condition and movement condition. As a result of this interaction, a follow-up repeated measures ANOVA for TA_{AVG} during the stationary trials for all three loading conditions was then conducted. A significant effect of condition on TA_{AVG} was found ($F(2,14) = 13.929$, $p < 0.001$, $\eta^2 = 0.499$, *Power* = 0.996). Bonferroni post hoc tests yielded a significant difference between the U (18.70°) and SDS (21.82°) conditions ($p = 0.004$) and the U (18.70°) and MDS (21.94°) conditions ($p = 0.009$). Both the SDS and MDS conditions yielded a larger TA_{AVG}, indicating that the participants were in a more flexed position during those two conditions, than during the unloaded condition. Although a significant effect of condition for TA_{AVG} was found ($F(2,14) = 13.929$, $p < 0.001$, $\eta^2 = 0.499$, *Power* = 0.996), there was no significant difference between the SDS (21.82°) and MDS (21.94°) conditions ($p = 0.824$), but the MDS condition did yield a larger TA_{AVG} which indicates that the participants were in a more flexed position during that condition (**Figure 31**).

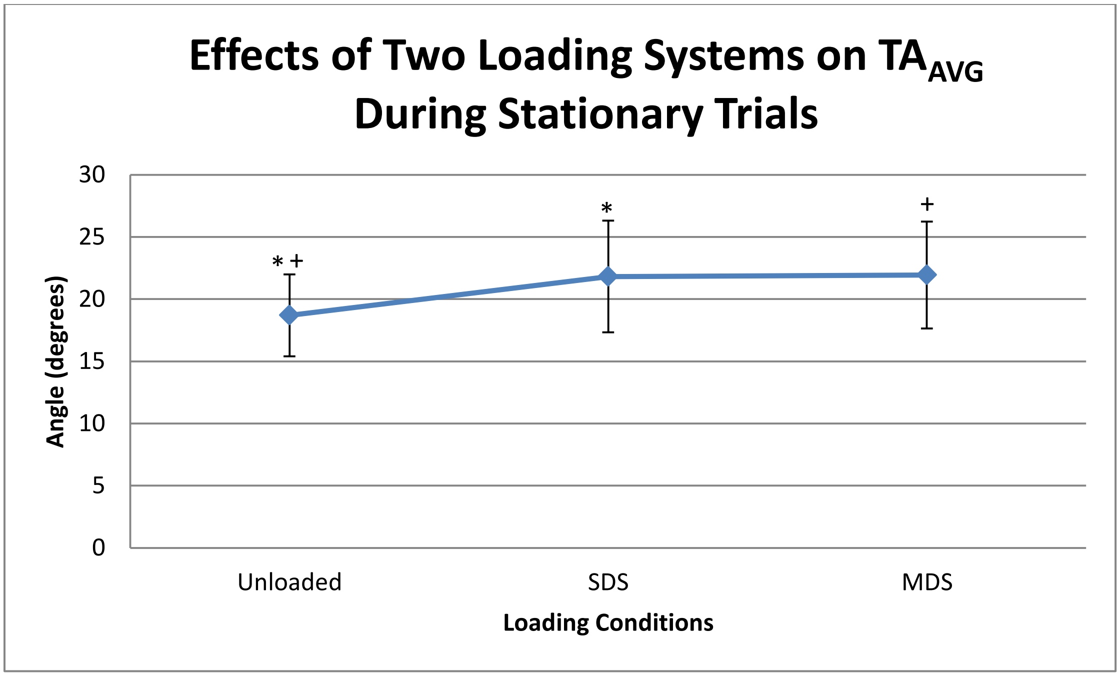


Figure 31. Effects of loading conditions on TA_{AVG} during stationary trials.
U-SDS: $*p = 0.004$. U-MDS: $^{+}p = 0.009$.

A follow-up repeated measures ANOVA for TA_{AVG} during the dynamic trials for all three loading conditions was conducted. A significant effect of loading condition on TA_{AVG} was found ($F(2,14) = 88.236$, $p < 0.001$, $\eta^2 = 0.855$, *Power* = 1.000). Bonferroni post hoc tests yielded a significant difference specifically between the U (15.79°) and SDS (22.96°) conditions ($p < 0.001$), and the U (15.79°) and MDS (22.71°) conditions ($p < 0.001$). Both the SDS and MDS conditions yielded a larger TA_{AVG}, indicating that the participants were experiencing more trunk flexion during those conditions than during the unloaded condition. Although an overall significant effect of loading condition on TA_{AVG} was found ($F(2,14) = 88.236$, $p < 0.001$, $\eta^2 = 0.855$, *Power* = 1.000), there was no significant difference between the SDS (22.96°) and MDS (22.71°) conditions (p = 1.000), it should be noted that the SDS condition did yield a larger TA_{AVG} which indicates that the participants were in a more flexed position during that condition (**Figure 32**).

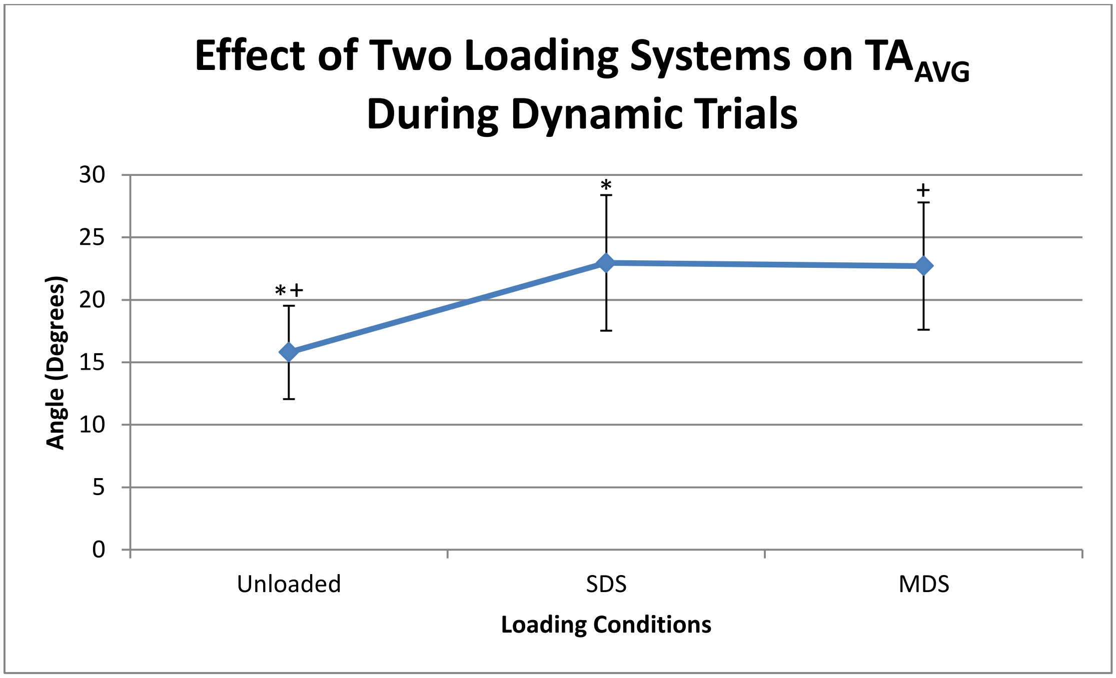


Figure 32. Effects of loading conditions on TA_{AVG} during dynamic trials.
U-SDS, U-MDS: $*^{+}p < 0.001$.

Average Hip Angle Over One Stride

The findings from the repeated measures MANOVA revealed that there was a significant interaction between loading condition and whether or not the participants were stationary or dynamic. As a result of this interaction, a follow-up repeated measures ANOVA for HA_{AVG} during the stationary trials for all three loading conditions was conducted. A significant effect of loading condition on HA_{AVG} found ($F(2,14) = 4.440$, $p = 0.020$, $\eta^2 = 0.228$, *Power* = 0.719). Bonferroni post hoc tests yielded a significant difference only between the U (73.82°) and SDS (77.08°) loading conditions ($p = 0.043$). There was no significant difference between the U (73.82°) and MDS (78.09°) loading conditions ($p = 0.057$), or between the SDS (77.08°) and MDS (78.09°) conditions ($p =$

1.000). These findings indicate that SDS produced a more vertical thigh position than the unloaded condition, but the MDS condition, though not significant, yielded the most vertical thigh position (**Figure 33**).

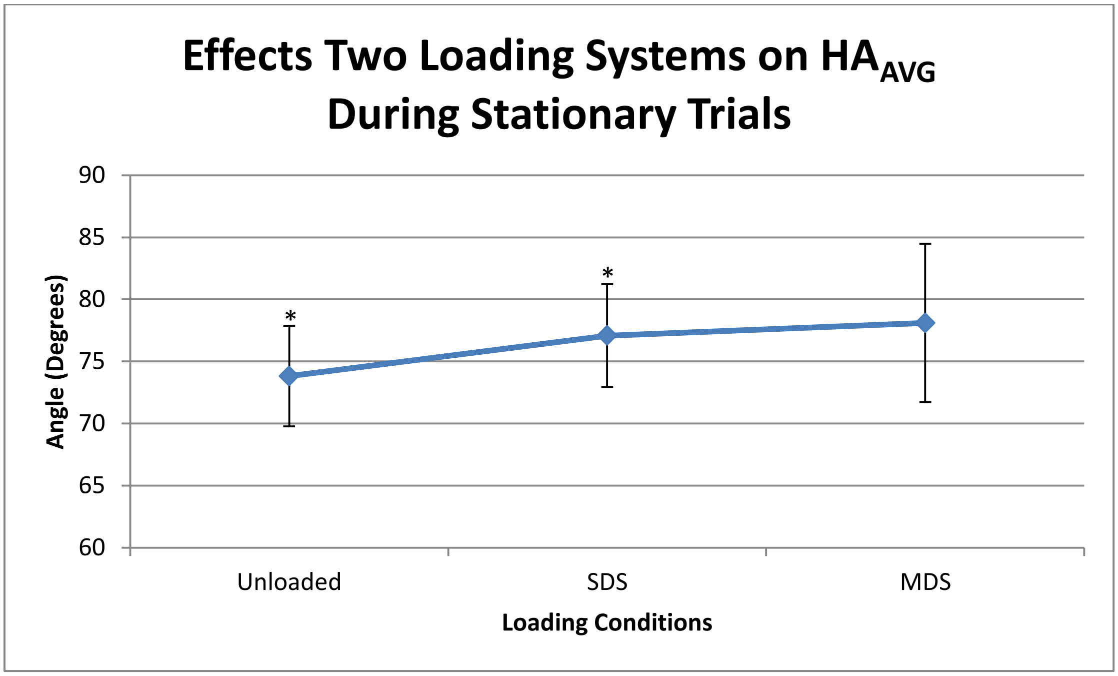


Figure 33. Effects of loading conditions on HA$_{AVG}$ during stationary trials.
U-SDS: *p = 0.043.

A follow-up repeated measures ANOVA for HA$_{AVG}$ during the dynamic trials for all three loading conditions was conducted. A significant effect of condition on HA$_{AVG}$ was found (F(2,14) = 4.638, p = 0.018, η^2 = 0.236, *Power* = 0.739). Bonferroni post hoc tests however did not yield a significant difference between any of the loading conditions: U (86.48°) and SDS (88.30°) conditions (p = 0.097), U (86.48°) and MDS (88.72°) conditions (p = 0.111), and SDS (88.30°) and MDS (88.72°) conditions (p = 1.000). Similar to the results found during the stationary movement condition, although there was no significant difference between the three loading conditions, the MDS condition yielded a larger HA$_{AVG}$, indicating less hip flexion angle (**Figure 34**).

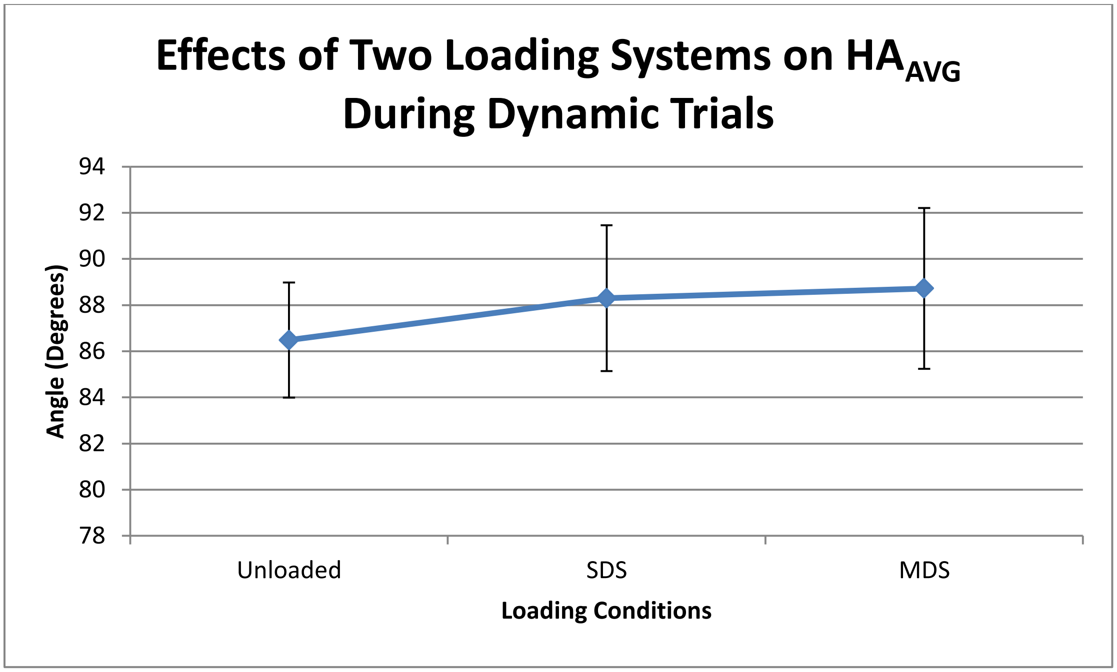


Figure 34. Effects of loading conditions on HA_{AVG} during dynamic trials.
U-SDS: $p = 0.097$. U-MDS: $p = 0.111$. SDS-MDS: $p = 1.000$.

Average Ankle Angle Over One Stride

The findings from the repeated measures MANOVA revealed that there was a significant interaction between loading condition and movement condition (i.e., stationary or dynamic). As a result of this interaction, a follow-up repeated measures ANOVA for AA_{AVG} during the stationary trials for all three loading conditions was conducted. A significant effect of loading condition on AA_{AVG} was found ($F(2,14) = 32.516$, $p < 0.001$, $\eta^2 = 0.684$, *Power* = 1.000). Post hoc tests using Bonferroni yielded a significant difference specifically between the U (91.35°) and SDS (93.75°) loading conditions ($p < 0.001$), and between the U (91.35°) and MDS (93.76°) loading conditions ($p < 0.001$). In both cases respectively, the SDS and MDS conditions yielded a larger AA_{AVG}, indicating that the ankle was in a more plantarflexed position during those conditions in comparison to the unloaded condition (**Figure 35**).

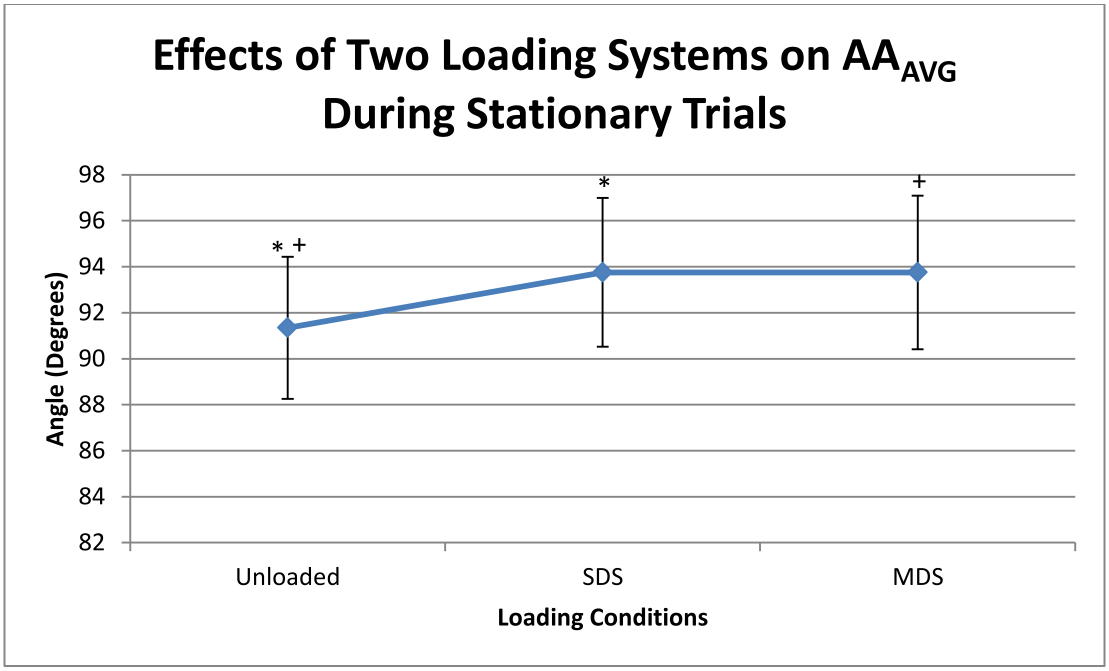


Figure 35. Effects of loading conditions on AA_{AVG} during stationary trials.
U-SDS, U-MDS: $*^{+}p < 0.001$.

A follow-up repeated measures ANOVA for AA_{AVG} during the dynamic trials for all three loading conditions was conducted. There was no significant effect of loading condition on AA_{AVG} found ($F(2,14) = 0.763$, $p = 0.475$, $\eta^2 = 0.048$, *Power* = 0.167) between the U (90.35°) and SDS (90.51°) loading conditions, between the U (90.35°) and MDS (89.80°) loading conditions, or between the SDS (90.51°) and MDS (89.80°) loading conditions. All three conditions yielded ankle angles very close to the regimented 90° required during a marching performance (**Figure 36**).

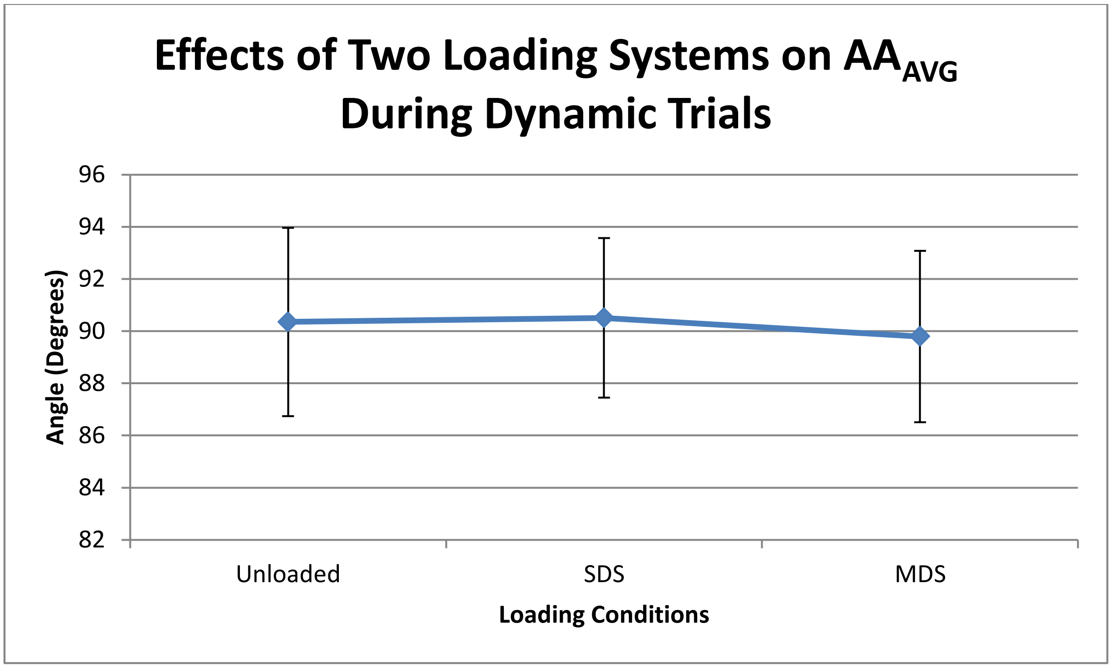


Figure 36. Effects of loading conditions on AA_{AVG} during dynamic trials.
U-SDS: p = 1.000. U-MDS: p = 0.923. SDS-MDS: p = 0.579.

Average Knee Angle During Foot Flat (FF)

In addition to investigating the effects of the SDS and MDS loading conditions on the joint kinematics of the body during a stride, the author also investigated the average joint kinematics of the knee during the FF phase of the stride length (KA_{AFF}). In order to determine significance, a 3 (condition: unloaded (U), SDS, MDS) x 2 (time: pre, post) multivariate repeated measures (ANOVA) across knee ankle with a p value set *apriori* at $p < 0.05$ was employed (Wilks's Λ = 0.928, F(2,14) = 0.545, p = 0 .592, η^2 = 0.072, *Power* = 0.122). The multivariate repeated measures ANOVA did not yield a significant effect for loading condition, (F(2,14) = 1.114, p = 0.341, η^2 = 0.069, *Power* = 0.227), nor time (pre versus post) on the KA_{AFF}, (F(2,14) = 0.125, p = 0.729, η^2 = 0.008, *Power* = 0.063): U Pre (176.45°), SDS Pre (175.8°), MDS Pre (175.17°), U Post (176.21°), SDS Post (175.54°), & MDS Post (175.47°). This indicates that the loading conditions do not alter joint kinematics differently during the foot flat phase of the gait cycle. Descriptive statistics for all dependent variables are presented in **Appendix D** (**Figure 37**).

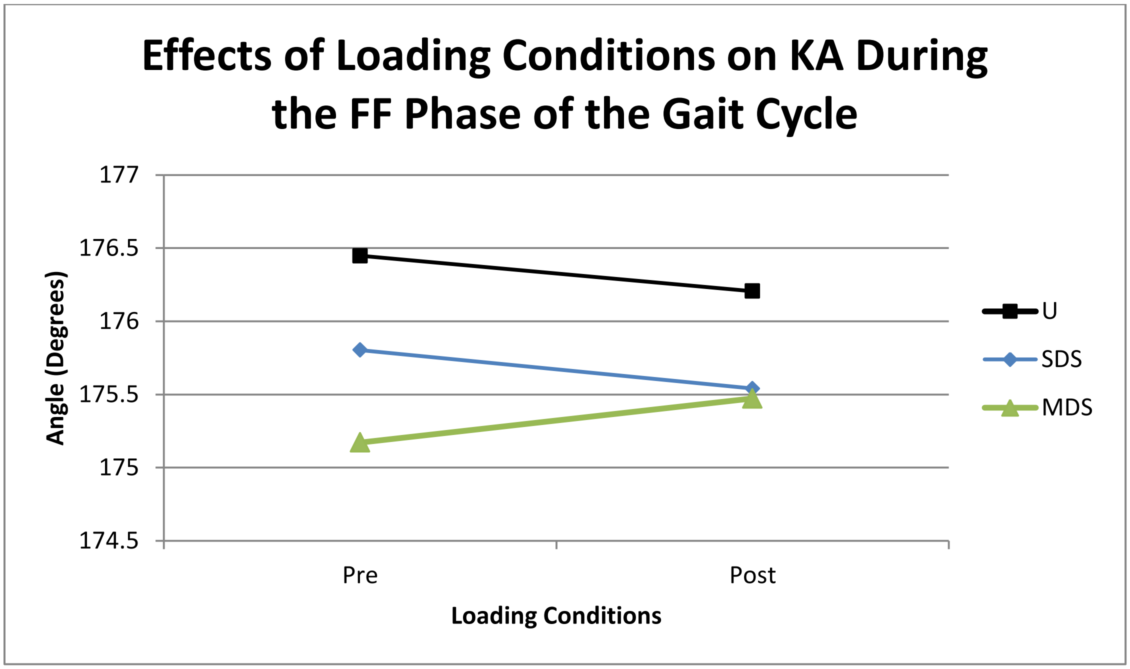


Figure 37. Effects of loading conditions on KA_{AFF}. p_{Load}= 0.341. p_{Time}= 0.729.

Average Ankle Angle During Foot Flat (FF)

In addition to investigating the effects of the SDS and MDS loading conditions on the joint kinematics of the body during a stride, the author also investigated the average joint kinematics of the ankle during the FF phase of the gait cycle (AA_{AFF}). In order to determine significance, a 3 (condition: unloaded (U), SDS, MDS) x 2 (time: pre, post) multivariate repeated measures (ANOVA) across knee ankle with a *p* value set *apriori* at $p < 0.05$ was employed (Wilks's $\Lambda = 0.917$, $F(2,14) = 0.630$, $p = 0$ .547, $\eta^2 = 0.083$, *Power* = 0.135). The multivariate repeated measures ANOVA did not yield a significant effect for loading condition (modified, SDS and MDS), ($F(2,14) = 0.467$, $p = 0.631$, $\eta^2 = 0.030$, *Power* = 0.119), nor time (pre versus post) on the AA_{AFF}, ($F(2,14) = 0.691$, $p = 0.419$, $\eta^2 = 0.044$, *Power* = 0.122): U Pre (87.92°), SDS Pre (87.51°), MDS Pre (87.53°), U Post (88.48°), SDS Post (88.56°), & MDS Post (87.40°). Similar to the kinematic results at the knee, this analysis indicates that loading conditions do not alter joint kinematics differently during the foot flat phase of the gait cycle. Descriptive statistics for all dependent variables are presented in **Appendix D** (**Figure 38**).

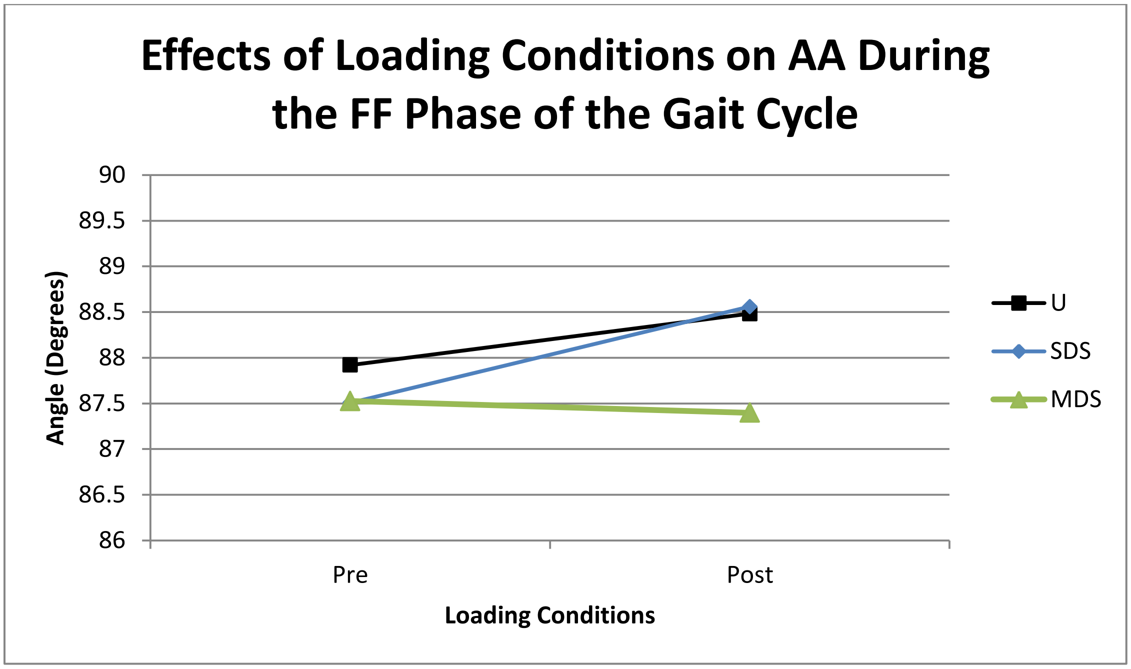


Figure 38. Effects of loading conditions on AA_{AFF}. $p_{Load} = 0.631$. $p_{Time} = 0.419$

Section 3: Erector Spinae Muscle Activity

In addition to examining the effects of an SDS and MDS system on joint kinematics, this research study also investigated the outcome of both the SDS and MDS systems on the muscular demands of the erector spinae muscle group (ES) while standing and playing in place (stationary) and marching forward while playing (dynamic). This section includes the results from the SDS and MDS loading conditions compared to the unloaded condition in terms of average (EMG_{AVG}) and peak EMG (EMG_{PEAK}) levels, including follow-up tests for specific conditions.

Average sEMG

A 2 (movement: stationary vs. dynamic) x 3 (loading condition: U, SDS and MDS) x 2(time: pre and post a one minute marching exercise) multivariate repeated measures ANOVA with a p value set *apriori* at $p < 0.05$ was employed and yielded a significant interaction effect of loading condition and time on the average muscle activity of the ES, ($F(2,14) = 1.006$, $p = 0.041$, $\eta^2 = 0.252$, *Power* = 0.618). This indicates that the muscle activity of the ES differed across the loading conditions, pre and post the one minute marching exercise. Descriptive statistics for all dependent variables are presented in **Appendix E**.

As a result of this interaction, a follow-up repeated measures ANOVA for EMG_{AVG} during the pre-trials for all three loading conditions was conducted, and a significant effect of loading condition on EMG_{AVG} was found ($F(2,14) = 28.958$, $p < 0.001$, $\eta^2 = 0.674$, *Power* = 1.000). Post hoc tests using Bonferroni yielded a significant difference between the U (1.70% MVIC) and SDS (3.57% MVIC) loading conditions ($p < 0.001$), and MDS (3.64% MVIC) ($p < 0.001$) loading condition with the SDS and MDS conditions yielding higher EMG_{AVG} levels than the unloaded condition. There was no significant difference between the SDS (3.57% MVIC) and MDS (3.64% MVIC) conditions ($p = 1.000$). This indicates that the participants' average ES muscle activity increased while wearing the SDS and MDS systems during both stationary and dynamic tasks, before participating in the one minute marching exercise (**Figure 39**).

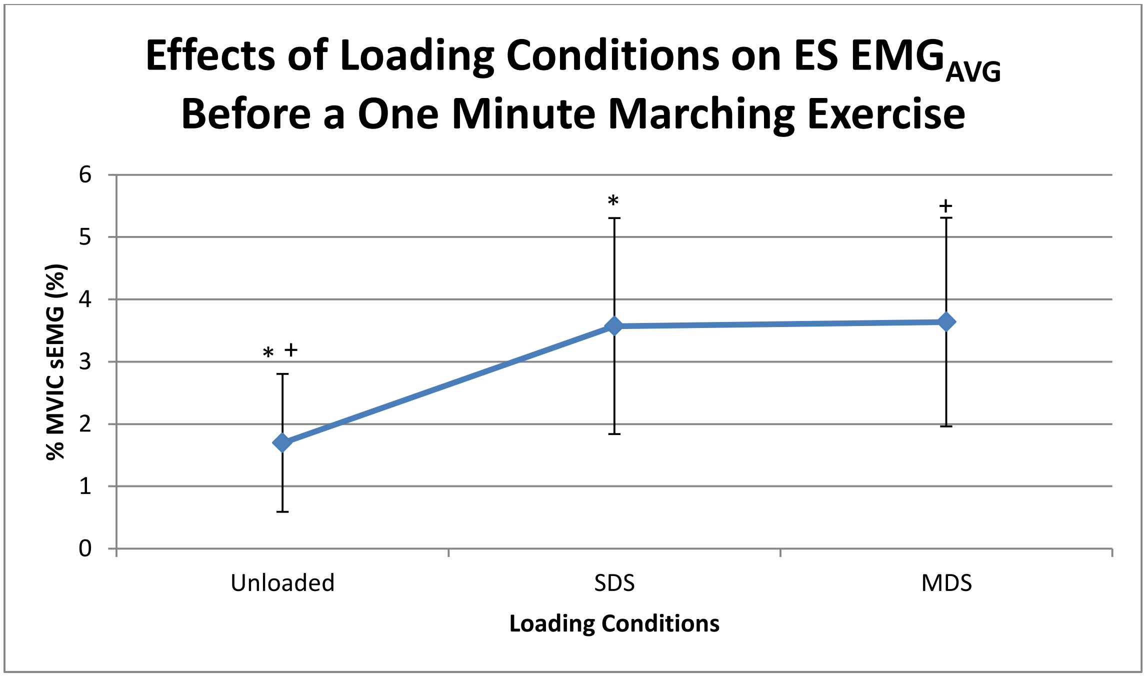


Figure 39. Effects of loading conditions on ES EMG_{AVG} before a one minute marching exercise. U-SDS, U-MDS: $*^{+}p < 0.001$.

A follow-up repeated measures ANOVA for EMG_{AVG} during the post trials for all loading conditions across the non-included variable of movement (stationary vs. dynamic) was conducted. A significant effect of loading condition on EMG_{AVG} was found ($F(2,14) = 29.426$, $p < 0.001$, $\eta^2 = 0.710$, *Power* = 1.000). Post hoc tests using Bonferroni yielded a significant difference between the U (1.72% MVIC) and SDS (4.01% MVIC) loading conditions ($p < 0.001$), and MDS (3.68% MVIC) with the SDS and MDS conditions yielding higher EMG_{AVG} level. There was no significant difference between the SDS (4.01% MVIC) and MDS (3.69% MVIC) conditions ($p = 0.558$). This indicates that the participants' average ES muscle activity was increased while wearing the SDS and MDS systems during both stationary and dynamic tasks, after participating in the one minute marching exercise (**Figure 40**).

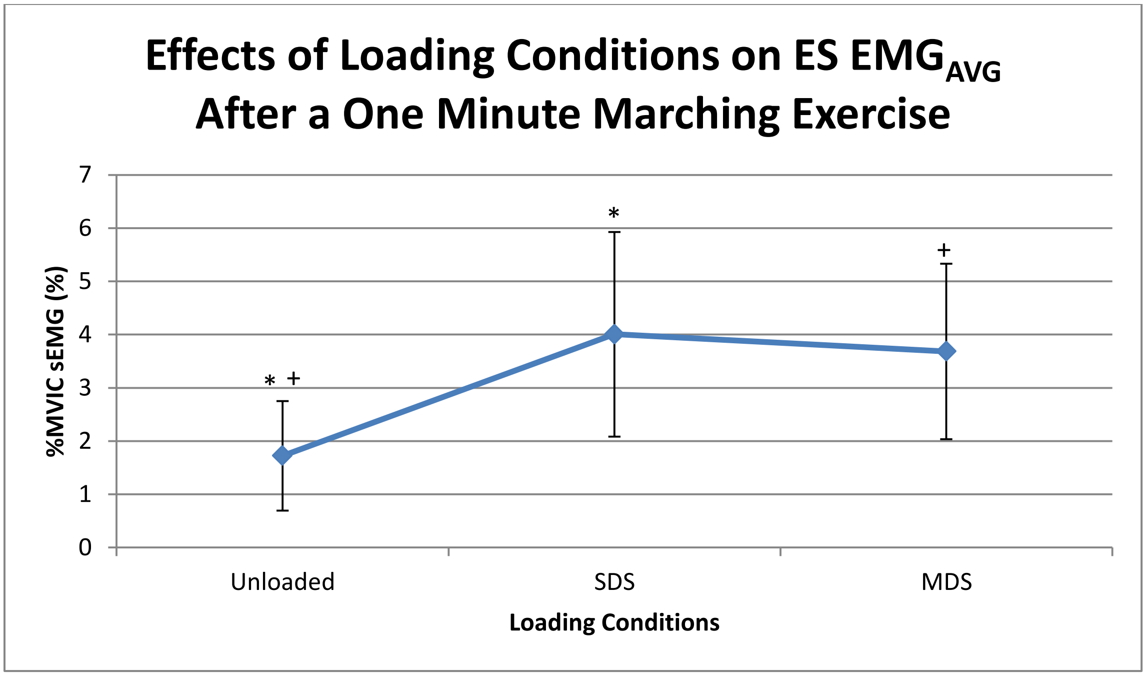


Figure 40. Effects of loading conditions on ES EMG_{AVG} after a one minute marching exercise. U-SDS: *$p < 0.001$.

Peak sEMG

A 2 (movement: stationary vs. dynamic) x 3 (loading condition: U, SDS and MDS) x 2(time: pre and post marching) a multivariate repeated measures ANOVA with a *p* value set *apriori* at $p < 0.05$ yielded a significant interaction effect of loading condition and movement on the peak muscle activity of the ES, ($F(2,14) = 6.996$, $p = 0.005$, $\eta^2 = 0.412$, *Power* = 0.883). This indicates that the loading conditions altered the muscle activity of the ES differently across movement conditions. Descriptive statistics for all dependent variables are presented in **Appendix E**.

The findings from the repeated measures ANOVA revealed that there was a significant interaction between loading condition and whether or not the participants were stationary or dynamic. As a result of this interaction, a follow-up repeated measures ANOVA for EMG_{PEAK} during the stationary trials for all loading conditions was conducted. A significant effect of loading condition on EMG_{PEAK} was found ($F(2,14) = 25.847$, $p < 0.001$, $\eta^2 = 0.665$, *Power* = 1.000). Post hoc tests using Bonferroni yielded a significant difference between the U (3.05% MVIC) and SDS (5.61% MVIC) loading conditions ($p < 0.001$), and the U (3.05% MVIC) and the MDS (5.30% MVIC) loading conditions ($p = 0.001$), with both the SDS and MDS conditions yielding higher EMG_{PEAK} levels. There was no significant difference between the SDS (5.61% MVIC) and MDS (5.30% MVIC) conditions ($p = 0.995$). This indicates that the participants' peak ES muscle activity increased and was affected when wearing the SDS and MDS systems during the stationary task (**Figure 41**).

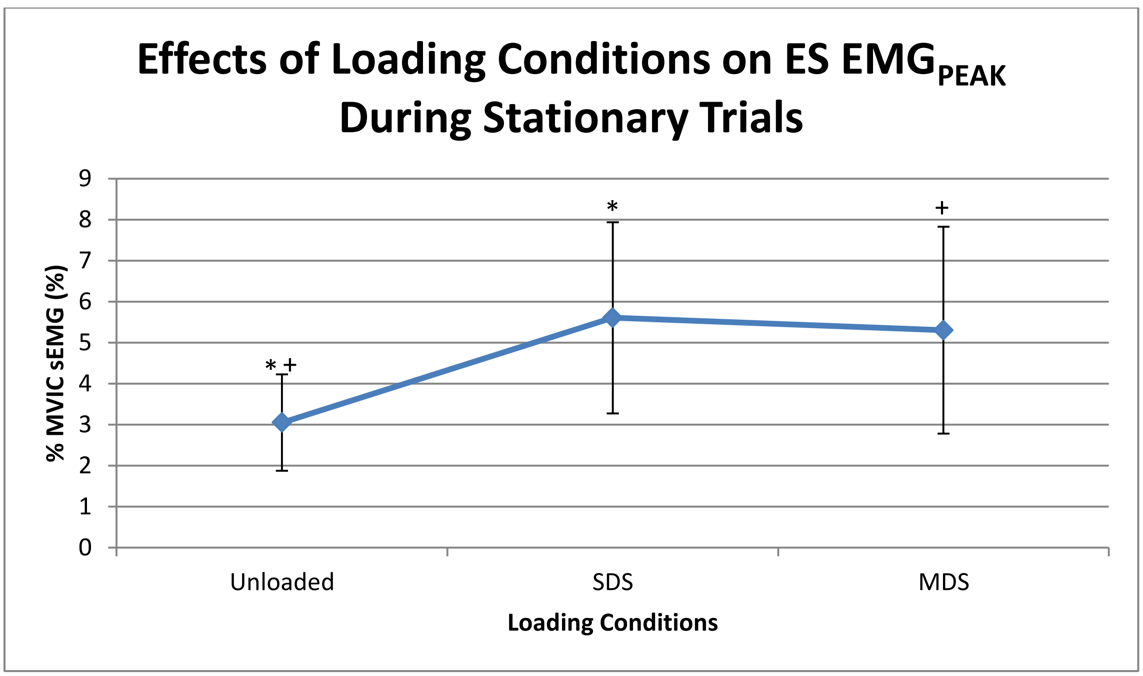


Figure 41. Effects of loading conditions on ES EMG_{PEAK} during stationary trials. U-SDS: *$p < 0.001$.

A follow-up repeated measures ANOVA for EMG_{PEAK} during the dynamic trials for all loading conditions was conducted. A significant effect of loading condition on

EMG_{PEAK} was found ($F(2,14) = 20.834$, $p < 0.001$, $\eta^2 = 0.654$, *Power* = 1.000). Post hoc tests using Bonferroni yielded a significant difference between the U (5.62% MVIC) loading condition and SDS (9.74% MVIC) ($p = 0.005$) and the U (5.62% MVIC) and MDS (10.25% MVIC) conditions ($p < 0.001$), with both the SDS and MDS conditions yielding higher EMG_{PEAK} levels. There was no significant difference between the SDS (9.74% MVIC) and MDS (10.25% MVIC) conditions ($p = 1.000$). This indicates that the participants' peak ES muscle activity increased and was affected when wearing both the SDS and MDS systems during the dynamic task (**Figure 42**).

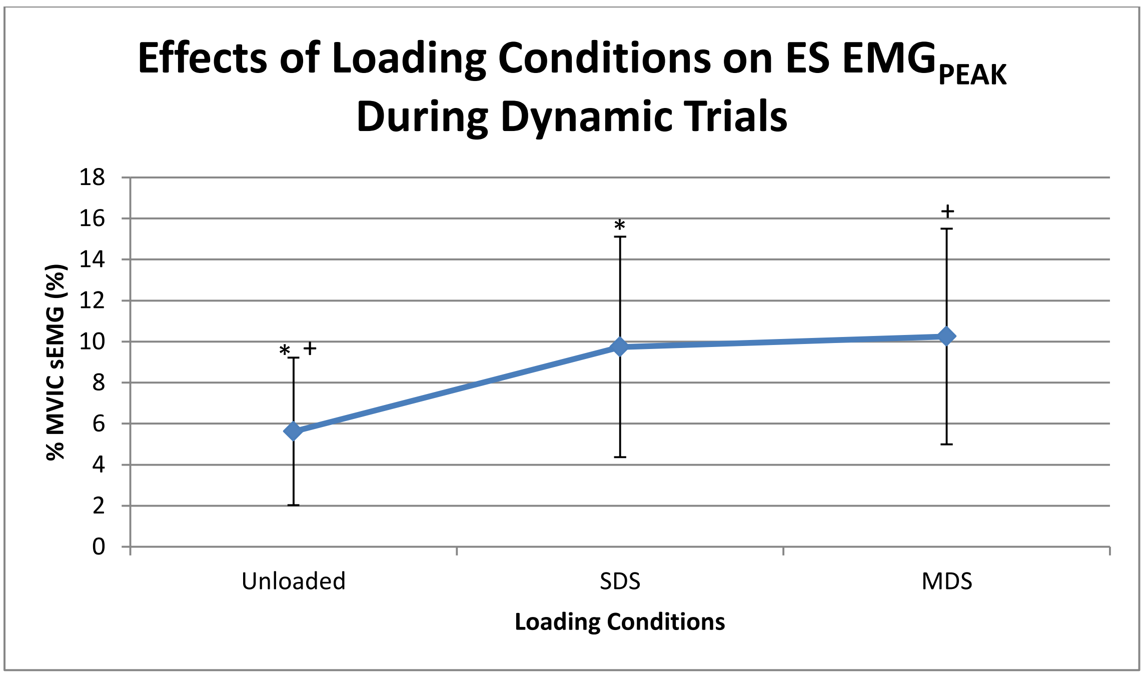


Figure 42. Effects of loading conditions on ES EMG_{PEAK} during dynamic trials. U-SDS: $*p = 0.006$. U-MDS: $^{+}p < 0.001$.

Section 4: Contact Pressure

In addition to kinematics and EMG, contact pressure at the abdominal area and shoulders were also examined. Contact pressures were investigated while in a stationary position for one minute while playing, with and without the presence of the lumbar belt (SDS vs. MDS), before and after a one minute marching exercise. These measures were taken at three times during the data collection: Pre5-during the initial 5 seconds of being loaded before the marching exercise, Pre55-during the last 5 seconds of the initial loading before the marching exercise, and Post55-during the last 5 seconds of marching in place after the one minute exercise.

A 2 (shoulder: right and left) X 3 (time: pre5, pre55, and post55) repeated measures ANOVA for Left Shoulder (LS_{PEAK}) and Right Shoulder (RS_{PEAK}) was conducted to determine if there was a significant difference between the contact pressure levels at both shoulder locations. Results revealed that there were no significant differences between the two locations, allowing the author to combine of the shoulder data into the peak contact pressure at the shoulder ($SHLDR_{PEAK}$): ($F(2,14) = 2.595$, $p = 0.128$, $\eta^2 = 0.147$, *Power* = 0.326).

Peak Shoulder and Abdominal Contact Pressures

Following the regression analysis, a 2 (loading condition: SDS and MDS) x 2 (body location: shoulder and abdominal) x 3(time: pre5, pre55 and post55) multivariate repeated measures ANOVA with a p value set *apriori* at $p < 0.05$ yielded a significant interaction of peak contact pressure between body location and loading condition, ($F(1,15) = 8.993$, $p = 0.009$, $\eta^2 = 0.375$, *Power* = 0.800). This indicates that the loading conditions altered the magnitude of contact pressures differently across the body locations, as well as where the load was distributed during the two loading conditions. Descriptive statistics for all dependent variables are presented in **Appendix F**.

In addition to a repeated measures ANOVA across the two loading conditions (SDS and MDS) per body placement, two paired *t-tests* were conducted to investigate both loading conditions (SDS and MDS) for $SHLDR_{PEAK}$ and ABS_{PEAK} respectively. Each test yielded a significant difference. For the $SHLDR_{PEAK}$ there was a significant difference between the SDS (11.59 kPa) and MDS (8.89 kPa) conditions (p = 0.021), with the MDS condition yielding lower pressures. For the ABS_{PEAK} there was a significant difference between the SDS (4.77 kPa) and MDS (5.48 kPa) conditions (p = 0.009), with the MDS condition yielding higher pressures than the SDS. These results indicated that placing the SDS load onto the participants' body resulted in the largest contact pressures at the shoulders but less pressure at the abdominal region due to the lumbar belt being implemented to redistribute the load to the abdominal plate, pelvis, and belt (**Figure 43**).

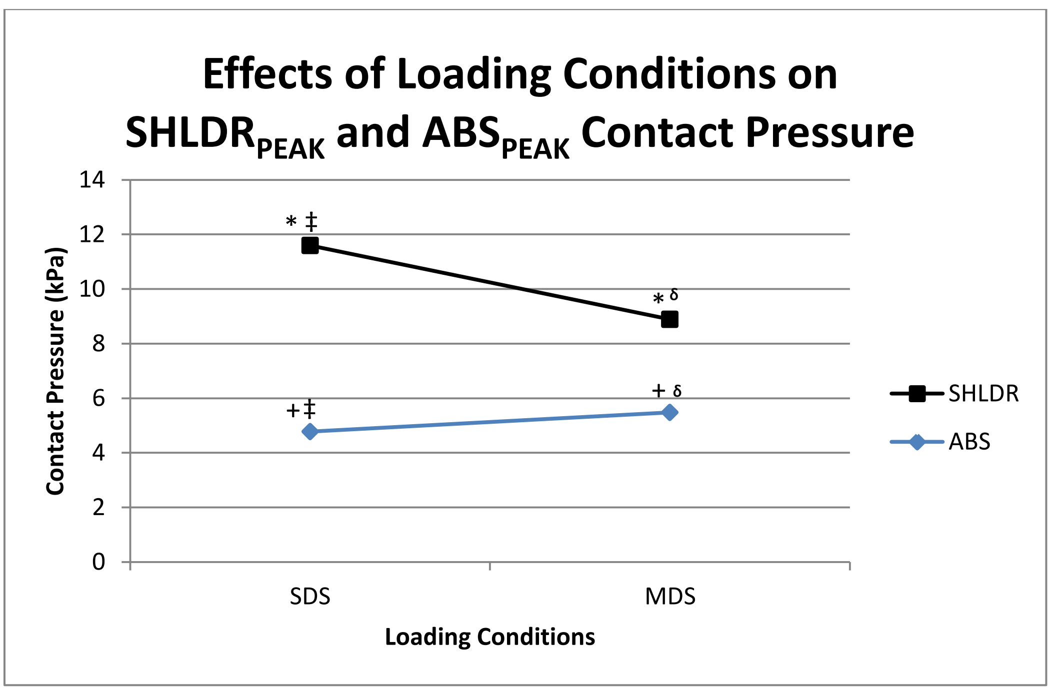


Figure 43. Effects of loading conditions on $SHLDR_{PEAK}$ and ABS_{PEAK} contact pressure. $^{*}p = .021$. $^{+}p = .009$. $^{‡}p = .001$. $^{\delta}p = .033$.

CHAPTER V

DISCUSSION

The purpose of this investigation was three-fold: 1) to determine the influence of a standard marching snare drum system (SDS) on trunk and lower extremity kinematics while standing and marching, 2) to determine the contact pressures and muscular demand placed on the erector spinae of individuals wearing a SDS, and 3) to investigate the influence of a lumbar belt (SDS-modified (MDS)) on trunk and lower extremity kinematics, muscular activity, and contact pressures. The following chapter has been divided into three primary sections: Section 1: The effect of the SDS and MDS systems on joint kinematics, Section 2: The effect of the SDS and MDS systems on ES muscle activity, and Section 3: Difference in contact pressures between the SDS and MDS loading conditions, at the shoulder and abdominal regions.

Section 4 includes final conclusions and recommendations for future research.

Section 1: SDS & MDS Conditions on Joint Kinematics

This section will discuss the results regarding the demands of wearing a SDS and MDS on the gait kinematics of marching drumline individuals, compared to an unloaded condition. The gait kinematics of interest are the average angular kinematics of the trunk (TA_{AVG}), hip (HA_{AVG}), and ankle (AA_{AVG}). It should be noted that knee angle was also investigated, but no significant differences were found for that particular joint, therefore will not be included in this discussion. The angular kinematics were investigated while the participants stood still and played the drum (stationary) as well as when they marched forward and played (dynamic). Table 4 illustrates the results for each joint and movement condition across all the two loading conditions.

Table 4
Effect of Loading Conditions on Joint Kinematics

Kinematic Variable	SDS	MDS
TA_{AVG} Stationary	* ↑	+ ↑
HA_{AVG} Stationary	* ↑	↑
AA_{AVG} Stationary	* ↑	+ ↑
TA_{AVG} Dynamic	* ↑	+ ↑
HA_{AVG} Dynamic	↑	↑
AA_{AVG} Dynamic	↑	↓

Note. ↑= Angle larger than U condition, (increased trunk flexion, increased ankle plantarflexion, and decreased hip flexion). ↓= Angle smaller than U condition (decreased trunk flexion, and increased ankle dorsiflexion). * or + = Significant difference from U condition

Previous literature has indicated that adaptations to an external load placed on the front of the body include trunk extension which shifts the COG_{sys} closer to an unloaded position (Devroey et al., 2007; Ling et al., 2004), and increased hip and knee flexion which improves shock absorption and lowers the COG_{sys} (Anderson et al., 2007; Cham & Redfern, 2004; Holbein & Redfern, 1997; Kinoshita, 1985). However, there are specific demands/constraints that are involved with participating in a marching band as a snare drummer that do not allow for most of these previously noted adaptations. The constraints include but are not limited to: restricted arm position, standardized step size, drum position, and body posture. As a result, individuals who wear a snare drum and harness will experience an anterior shift of the center of gravity of the person (COG_{person}), but will be unable to take advantage of the adaptations of trunk extension, natural arm swing, changes in stride length and frequency, nor changes in joint angles in an attempt to alleviate the stress being placed on the body. In observing a marching band practice or performance, drumline members seem to be maintaining an upright posture, however the trunk measure findings during both load conditions indicate that the drum caused the trunk to be in a significantly flexed position (See Figures 31 and 32). According to Keyserling (1986), neutral trunk position occurs at less than 20° from the vertical. In the present study, during both the stationary and dynamic movement conditions, the participants' unloaded trunk angles were within those parameters (18.70° and 15.79° of flexion respectively). However, during both the SDS and MDS conditions the TA_{AVG} was larger than 20°. This finding indicates that the external load was great enough to alter the trunk kinematics outside what has been considered neutral posture. This finding was surprising as it was anticipated that the anterior load would cause the performer to extend the trunk in order to bring the load closer to the base of support. In addition, it was thought that the addition of the lumbar support would attenuate the trunk extension. At this point it is unclear whether the trunk flexion is perceived as more aesthetically (determined by the director) pleasing or is an attempt by the performer to utilize the upper posterior portion of the harness support.

There were no significant changes in the hip during the dynamic movement condition; however, the trends did show that the MDS loading condition had the largest absolute hip flexion angle. Therefore it can only be speculated that the MDS condition allowed for increased ROM (flexion and extension at the hip) and, thus the larger HA_{AVG}. The increased hip flexion during the dynamic movement could also be an indication of the hip/upper leg being called upon to act as a shock absorber during the load carriage conditions. It has been stated in previous literature that the lower extremities will act as shock absorbers during load carriage (Kinoshita, 1985), and normally one would expect that the knee would be the most likely joint for this role. However, in the present study there were no significant changes in knee angle while stationary, throughout one stride (dynamic), nor during the loading phase (foot flat). This leads the author to conclude that the role of shock absorber for the body is unknown. It is reasonable to conclude that the lack of changes noted at the knee is the result of the restrictive nature of the marching gait which requires the participants to have very little knee flexion when making contact with the ground and to remain as upright as possible (i.e. no opportunity to lower the COG_{sys} closer to the ground) throughout the stance phase.

There was however, a significant difference in HA_{AVG} between the U and SDS conditions during the stationary condition. The SDS load resulted in a significantly larger HA_{AVG} which is indicative of a thigh position closer to vertical and in this movement condition indicates a more upright posture. This does not indicate that wearing the SDS causes better posture, but instead suggests to the author that the anterior shift in the COG_{person} due to the SDS system resulted in the participants leaning back, thus increasing the angle of the thigh to parallel. Without a significant change in the knee angle, future research is needed to determine if other factors such as pelvic girdle movement could be influencing the hip angle during stationary marching tasks.

The effect of loading an individual with either the SDS or MDS systems resulted in significant changes in AA_{AVG}, but only during the stationary condition. It was initially hypothesized that the amount of plantarflexion would increase during the stationary condition when loaded because of the shift in COG_{sys} forward, resulting in the participants attempting to shift the COG_{sys} back and doing so by pivoting about the ankle.

It was believed that while marching, the participants would also have an increase in dorsiflexion. The results however, did not find a significant effect of load condition for AA_{AVG}, but instead found that all three loading conditions had AA_{AVG} close to the fundamental 90° angle. This indicates that regardless of loading condition, when asked to perform the roll-step marching gait, that experienced marchers will automatically place the ankle at the required 90°.

Average Knee Angle During Foot Flat (FF)

In addition to investigating the effects of the SDS and MDS loading conditions on the joint kinematics of the body during a stride, the author also investigated the average joint kinematics of the knee during the FF phase of the stride length (KA_{AFF}). Results for this portion of the study did not yield a significant effect for loading condition (U, SDS, and MDS) nor time (pre, post). This indicates that the loading conditions do not alter the joint kinematics of the knee differently during the foot flat phase of the gait cycle.

Although not stated in the primary or secondary hypotheses, based on the current literature, it was expected that the SDS condition would cause an increase in knee flexion during the loading phase of the gait cycle. This posture would serve as a protective mechanism by the participants by lowering the COG_{person} for increased balance and control (Holt et al., 2003; Kinoshita, 1985). The lack of a finding supports the notion that the marching gait constraints supersede the load carriage demands.

Average Ankle Angle During Foot Flat (FF)

In addition to investigating the effects of the SDS and MDS loading conditions on the joint kinematics of the body during a stride, the author also investigated the average joint kinematics of the ankle during the FF phase of the gait cycle (AA_{AFF}). Results for this portion of the study were similar to the KA_{AFF} data, where no significant effect for loading condition (U, SDS, and MDS) or time (pre, post) was found. This indicates that the ankle kinematics were the same for all three loading conditions during the foot flat phase of the gait cycle. The reason for investigating this factor stemmed from the relationship between the lower leg and the foot, where it was believed that the SDS condition would yield more knee flexion and smaller ankle angle, indicating that the knee and ankle structures were acting as shock absorbers during the gait cycle. However, the constraints of the marching gait prevented such adaptation.

Section 2: SDS & MDS Condition on ES Muscle Activity

In addition to significantly affecting the joint kinematics of marching drumline members, load carriage tasks also affect the muscular activity of muscles involved in postural stability (i.e. ES). Therefore, the demands of wearing a SDS and MDS on the muscular activity of the ES while standing and playing in place (stationary) and marching forward while playing (dynamic) were also investigated. The average and peak EMG levels were compared across all three loading conditions. The previous section discussed the influence of load, movement, and time on the joint kinematics of the body, and it is those changes in joint kinematics (whether due to the testing variables or the restrictive nature of marching) that influence the demands on the muscles. Anderson et al. (2007) investigated the influence of carrying an anterior load on the body on the muscle activity of the ES, while participants stood still and walked. While holding the various loads anterior to the trunk, the researchers documented increased trunk extension and subsequent increases in ES muscle activity. This supported the development of the hypothesis that placing a load on the anterior portion of the body would result in increased muscle activity of the ES in order to maintain the erect body posture required by the demands of this type of performance. Table 4 illustrates all of the significant findings for the ES during each movement condition across all three loading conditions.

Table 5

Effect of Loading Conditions on sEMG of the ES

sEMG-ES Activity	SDS	MDS
EMG_{AVG} Pre	* ↑	+ ↑
EMG_{AVG} Post	* ↑	+ ↑
EMG_{PEAK} Stationary	* ↑	+ ↑
EMG_{PEAK} Dynamic	* ↑	+ ↑

Note. ↑ = Muscle activity larger than U condition. *or $^{+}$ = Significant difference from U condition

It can be concluded from the present project that load carriage, particularly either the SDS or MDS systems, affect levels of EMG_{AVG} and EMG_{PEAK} ES activity in collegiate male and female marching drumline members. This supports the finding of Bobet and Norman (1984) and Hamilton et al. (2008), which indicated that deviations from a normal, unloaded, anatomically neutral posture can result in a change in muscular demand. An additional demand that is placed upon the postural musculature is the inability to utilize arm movement for balance control, due to performance constraints. Natural arm movements modulate the excursions of the COG_{sys} (Birrell & Haslam, 2008), and therefore, when those movements are limited or restricted, the result can be greater excursions of the COG_{sys} within the BOS and therefore decreased stability. In order to compensate for the instability, stabilizing muscle activity (i.e. postural muscles) will increase, resulting in an earlier onset of fatigue or "fatigue-like effects" (Cook & Neumann, 1987).

Al-Khabbaz et al. (2008) determined that placing a load on the posterior part of the body (10, 15, and 20% BM) would result in an increase in muscle activity (ranging from 3-6% MVIC) of the rectus abdominus (RA) in an attempt to counteract the eccentric trunk extension moment. Motmans et al. (2006) also found that during a stationary posterior load carriage condition, the muscle activity of the ES decreased while the RA increased as a result of the extension moment. Interesting to note, even though Al-Khabbaz et al. (2008) determined that the RA activity had increased to compensate for the posterior load and backward inclination, the trunk was still in an extended position. Similar to the findings of Al-Khabbaz et al. (2008), Foti, Davids & Bagley (2000) and Motmans et al. (2006), the present study significant increases in muscle activity in the presence of a change in trunk position. For this particular study, it was a significant change in ES activity for the change in trunk flexed position.

For the current study, EMG_{AVG} was analyzed before (results ranging from 1.7-3.6% MVIC) and after a one-minute exercise (results ranging from 1.72-4.01% MVIC), similar to the findings of Al-Khabbaz and colleagues (2008) of 2.96-5.37% MVIC for 0-15% BM loads. The researchers believed that this increase in muscle activity was an attempt to counteract the eccentric trunk flexion moment that was occurring due to the anterior loading conditions. Similar results were found for the EMG_{PEAK} where during the stationary trials (results ranging from 3.05-5.61% MVIC) and the dynamic trials (results ranging from 5.62-10.25% MVIC), significant differences were found between the U and both SDS and MDS conditions. Although these levels of muscle activity were not the same amount of activity to the findings of Anderson et al. (2007) (results ranging from 30s-40s% MVIC), the data do indicate that there is a significant difference in muscle activity when there has been a change in postural demand due to an external load. Overall, the effect of placing an external load on the body results in postural demands as the COG_{sys} shifts anteriorly. Changes in posture result in changes in muscle activity, specifically the ES attempting to shift the COG_{sys} more within the BOS by extending the trunk. During both the stationary and dynamic trials, before and after the marching exercise, there was a significant difference in muscle activity of the ES for both the SDS and MDS conditions from an unloaded position. Therefore, it is reasonable to conclude that the participants' postures was influences by being loaded with a snare drum and carrier, and that before the marching exercise, the participants were not utilizing the load distribution characteristics of the lumbar belt. After the marching exercise, there were still significant differences for the stationary and dynamic tasks for both the SDS and MDS conditions and no significant difference between the two loads. However, there was a trend of decreasing muscular demand during the MDS to indicate that the participants might be making subtle changes in posture, therefore allowing the participants to utilize the load distribution characteristics of the lumbar belt.

Section 3: SDS & MDS Condition on Contact Pressure

The purpose of this portion of the current study is to present the findings and implications of peak contact pressures at the shoulders and abdominal region. These data were collected while wearing a SDS and MDS in a stationary position for one minute while playing, prior to and after a one minute marching exercise. The pressure data were taken at three times: Pre5-during the initial 5 seconds of the stationary/playing loading one minute period prior to the one minute marching exercise, Pre55-during the last 5 seconds of the stationary/playing loading one minute period prior to the one minute marching exercise, and Post55-during the last 5 seconds of the stationary/playing one minute period after the one minute exercise. The data of interest included the peak contact pressure values from both loading conditions (SDS and MDS), at both the shoulders ($SHLDR_{PEAK}$) and abdomen (ABS_{PEAK}). Table 5 illustrates the significant findings for the contact pressures for each loading condition, across both body segments.

Table 6
Effect of Loading Conditions on Contact Pressure

Contact Pressure	SDS	MDS
$SHLDR_{PEAK}$	⬆	* ⬆
ABS_{PEAK}	⬆	* ⬆

Note. ⬆ = Highest contact pressure. * = Significant difference between SDS and MDS.

Skin contact pressure is considered one of the primary limiting factors in the ability to perform a load carriage task (Holewijn, 1990), particularly if the load is not removed or reduced after a period of time. Doan (1998) determined that levels as low as 14 kPa can be a limiting factor, even though the participants in that study were not reporting any discomfort until 18-20 kPa. Lower pressures however do not necessarily mean that damage is not occurring in the tissues or nervous system (Goodson, 1981), nor should individuals avoid acknowledging the discomfort. The current study collected $SHLDR_{PEAK}$ levels between the SDS (11.59 kPa) and MDS (8.89 kPa) conditions, and ABS_{PEAK} levels between the SDS (4.77 kPa) and MDS (5.48 kPa) conditions, which are lower than the limiting value of 14 kPa determined by Doan (1998) and the performance criterion of 20 kPa indicated by Stevenson et al., 2004.

The present study also found a statistically significant decrease in pressure between the SDS and MDS configurations, similar to Holewijn (1990) and Mackie et al. (2005) who noted that implementing a lumbar belt significantly changes the distribution and amount of pressure being applied to the body during a load carriage condition.

Adding a lumbar belt to the SDS system not only reduced the amount of pressure at the shoulders, but also redistributed the forces to the abdominal region and pelvis which are areas better suited for load carriage (Scribano, 1970) (Figure 43). Anecdotally, participants remarked about the discomfort of the SDS system, but not the MDS system. Specific comments from the participants in a post data collection interview included: "Alleviates the pressure between the shoulder blades," "Takes away the pressure off the lower back right away," "Belt takes the weight off shoulders and off of you," "Doesn't push into stomach, just sits against it," and "Takes the pressure/weight off shoulders and puts on back and abs where the belt which feels good."

Overall Impact

Loading an individual with an anterior load, even a participant with experience in that particular loading condition, causes significant changes in joint kinematics, muscle activity, and contact pressure and the inclusion of a lumbar belt does not diminish the changes in kinematics and muscle activity. However, significant differences were noted in contact pressure findings. Implementing a lumbar belt did redistribute the contact pressures, changed the location of the drum with respect to the drummer, and most importantly did not harm the participants but instead made wearing a drum "feel better."

It was also discussed that the inclusion of the lumbar belt would decrease independent motion of the drum. One of the objectives of the lumbar belt was to make the drum and harness more connected to the performer and therefore decreases the drum from behaving as an independent inertial body. As an independent inertial body, the drum becomes like a person in a car without a seatbelt, the lumbar belt was thought to be the “seatbelt”. While this objective was difficult to measure quantitatively, the participants provided the following comments: "Cohesive," "Keeps the drum from bouncing around," "More aware of where the load is and how the drum moves, ""Now I have to focus on really rolling my feet properly so that the drum doesn't bounce," and "The belt brought the drum closer to the body, made it easier to move, and easier to perform maneuvers."

Section 4: Summary & Future Research

This final section will discuss the conclusions drawn from the current study and directions for future research. The goal of this study was investigate the loading effects of a traditional marching snare drum and carrier system on the kinematics, muscle activity, and contact pressures of the body. In addition to the baseline study, the author also investigated implementation strategies demonstrated by previous literature to alleviate the stress of load carriage on the body. The end result was the creation and construction of a lumbar belt specifically designed for marching drumline carriers, which was tested to determine changes in kinematics, muscle activity and contact pressures on the body.

When an individual wears a standard marching snare drum system while standing still, there are several significant changes: increased trunk flexion; decreased hip flexion; increased ankle plantarflexion (most likely as a result of the ankle acting as the axis of rotation for the large lever arm of the whole body and attempting to shift the COG_{person} posteriorly); increased ES EMG_{AVG} and EMG_{PEAK} as a result of the changes in trunk angle; and increased contact pressures at the shoulders indicating that the load is still predominately carried at the shoulders.

When an individual wears a standard marching snare drum system while marching, the results showed significant: increased trunk flexion, and increased ES EMG_{AVG} and EMG_{PEAK} as a result of the changes in trunk angle.

When implementing a lumbar belt there were similar significant findings at the joints and muscle activity level from the unloaded condition only, suggesting that the implementation of the lumbar belt did not alter the marching kinematics of the performer. However, a significant difference was noted between the two systems for contact pressure. It was determined that the MDS system caused lower contact pressure values at the shoulders and higher pressures at the abdominal region, thus indicating a redistribution of pressure and load to the belt, hips, abdomen, and pelvic girdle and off of the shoulders. Therefore it can be concluded that implementing a lumbar belt to a SDS system ameliorates skin contact pressure and discomfort at the shoulders, potentially decreasing the occurrence of an injury, and anecdotally providing relief to the participants.

Future Research

When considering load carriage, the primary variable of concern is the size of the load. Researchers have found significant changes in gait kinematics, muscle activity, and contact pressure with load sizes as low at 10% body mass (BM), and larger and more severe changes occurring at 20% BM. These findings have caused the literature to recommend that load carriage be limited to 20% BM (Devroey et al., 2007; Kinoshita, 1985; Malhotra & Sen Gupta, 1965; Pascoe & Pascoe, 1999). The results of this study found that the SDS system was on average 11.1% BM (ranging from 5.8-15.1% BM of the participants), explaining the significant findings of this study, while not exceeding the recommended load carriage limit of 20% for this particular population. However, the current study only investigated the effects of an SDS on collegiate marching band members. Yet this SDS is utilized by thousands of high school and junior high marching band participants and it is quite possible that this load could reach or exceed the recommended limit of 20% BM. Therefore, future research should be conducted on these younger populations to determine the effect of the SDS and MDS on these smaller performers.

Significant increases in muscle activity were found when Hong et al. (2008) investigated the effects of not only load carriage mass but load carriage duration. It was determined that the larger the mass the sooner it took for the muscle activity to reach a significantly different and fatigued level (i.e. 5 minutes at 20% BM, and 15 minutes at 15% BM respectively) . The EMG_{AVG} data from the present study demonstrated increases in muscle activity levels from pre to post a one minute marching exercise. Therefore variable of time (i.e. duration of wearing the SDS) will need to be addressed in future studies to determine duration effects of load carriage, adaptations or any learning effects, with the potential to identify efficient practice protocols.

In addition to these areas of future research, investigations should also be expanded to include 1) the other anterior load carriage scenarios in marching bands (i.e. quad/tenor drums, base drums, and cymbals), 2) vertical/direct loading to the spine in marching band (i.e. tubas), and 3) different populations of marching band participants (i.e. junior high, high school, and drum corps). As well as the long term effects (i.e. longitudinal studies) on participants in marching drumlines, based on anecdotal evidence of recurring injuries, pain, and discomfort attributed to being a marching drumline participant. These individuals, after all, are as passionate about this activity as an athlete is about their sport. However, less attention has been paid to the side effects of participation in marching activities compared to traditional athletic endeavors.

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Appendix A

Preliminary Medical Questionnaire

Please read each question carefully and answer honestly. If you do not understand the question, please ask the investigator for clarification. Check the appropriate answer.

Participant Number: ___________

YES **NO**

_____ _____ 1) Are you under the age of 19?

_____ _____ 2) Have you ever been told you have an inner ear disorder?

_____ _____ 3) Have you ever had lower extremity surgery?

_____ _____ 4) Do you presently have any lower extremity disorders?

_____ _____ 5) Has your doctor ever told you that you have a muscle, bone or joint problem such as arthritis that has been aggravated by exercise, or might be made worse by exercise?

_____ _____ 6) Have you ever felt faint, dizzy or passed out during or after exercise?

_____ _____ 7) Have you ever felt pain, pressure, heaviness or tightness in the chest, neck shoulders or jaws as a result of exercise?

_____ _____ 8) Do you have any reason to believe that your participation in this investigative effort may put your health or well-being at risk?

_____ _____ 9) Have you had an injury or surgery to either of your lower extremities?

_____ _____ 10) Are you able to walk and stand on one leg without the use of aids?

_____ _____ 11) Are you currently taking any medication that you think might influence your ability to participate in this study?

Appendix B

Institutionally Approved Informed Consent Document

INFORMED CONSENT
for a Research Study entitled

The Influence of a Marching Snare Drum Harness on Joint Kinematics, Electromyography and Contact Pressure

You are invited to participate in a research study to look at the load carriage demands placed on the collegiate marching band athlete. This study is being conducted Andrea Sumner and the supervision of Dr. Wendi Weimar, Associate Professor of Biomechanics, in the Auburn University Department of Kinesiology. You were selected as a potential participant because you are between the ages of 19 and 25 years of age, and your health condition and mobility might, through pre-screening health questionnaire to follow, permit you to perform these tests safely and successfully. The subject population is that of convenience and who represent the current collegiate marching band athlete. The results of this study will be used to investigate the biomechanical effects on the trunk and lower extremities associated with wearing a snare drum carrier and drum, and how the implementation of a lumbar belt to the snare drum carrier influences these effects.

What will be involved if you participate? If you decided to participate in this experiment, your testing session will last approximately 45 minutes, which will include time to fill out forms. After meeting the criteria for participation, both your mass and height will be measured for normalizing purposes. You will then be asked to change into spandex shorts and spandex tank tops to minimize clothing artifact. In order to identify the muscle group, the researchers will ask you to perform a trunk extension (bending backwards) movement. After the joint movement, the skin area will be cleaned with a 70% alcohol solution and abraded to reduce the electrical impedance of the skin. A ground electrode will be placed on the bony aspect of the olecranon process (elbow) and two additional electrodes will be placed on the muscles at your lumbar 4-5 interspace. Next, MVIC (maximum voluntary isometric contraction-used for normalization) will be collected for the group.

Following the sEMG (electromyography-muscle activity) electrode placement, three pressure sensors will be laid over the designated landmarks. Two pressure sensors (#9811E) will be placed directly under the location where the drum carrier makes contact with the body. The sensors will lie over the superior (top) aspect of the shoulders, angled towards the midline, and cross over the clavicle. The third sensor (#9833E) sensor will be placed across the lower abdominal/pelvic region, directly under the location where the drum carrier makes contact with the front of the body with the center of the sensor at the midline of the body.

Finally, retroreflective markers will be affixed to the spandex clothing and to the skin with double-sided tape and placed according to the requirements of the Vicon® Plug-In-Gait© model. A static capture will be performed while you stand in the center of the capture volume in an anatomical position with shoulders out to the side. Once the static capture is performed, you will begin the data collection. The procedures for this study are the same throughout the study, with the only changes occurring for the different loading conditions. The loading conditions consist of unloaded, wearing your snare drum and harness without any modifications, and wearing your snare drum and harness with the addition of a lumbar belt. The procedures for this study for each loading condition are as follows: 1) you will be asked to stand for 1 minute while sEMG, pressure data, and joint kinematics are collected, 2) you will then march (glide step and crab step gait) through capture volume while sEMG, pressure, and joint/gait kinematic data is collected, 3) you will then march (glide and crab step gait) for 10 minutes in the coliseum gymnasium, 4) you will then reenter the capture volume and stand in a stationary position for 1 minute while EMG, pressure data, and joint kinematics will be collected once again, and 5) you will then march (glide step and crab step gait) through capture volume while sEMG, pressure, and joint/gait kinematic data is collected.

During the static trials, your arms will be in the playing position and your feet will be at a 45° angle. During the unloaded condition, the arms will be in the playing position as well in order for control for arm placement throughout the conditions, as well as to mimic the requirements that occur during practice of "air playing." There will be a period of 5 minutes of rest in between each loading condition, and more will be afforded to you if needed.

Are there any risks of discomforts? There is a slight risk associated with walking, marching, and carrying an instrument. However, since these are usual activities for you as a collegiate marching band member, these risks are minimal. Last, a Certified First Aid/First Responder will be present at all testing sessions. If medical treatment is necessary due to an injury, you are responsible for any and all medical costs.

Are there any benefits to yourself or others? As a participant, it is hoped that you will have a better understanding as to how load carriage effects your gait (if it does), and how the demands placed on your body is affected by the current harness system you have and one that had been modified. Hopefully this information will provide the researchers with ideas that can help you as an athlete to be more efficient on the field and less susceptible to injuries. We/I cannot promise you will receive any of the benefits described.

Will you receive compensation for participating? To thank you for your time, you may request the data. There will be no compensation for participating in this project.

If you change your mind about participating, you can withdraw at any time during the study. Your participation is completely voluntary. If you choose to withdraw, your data can be withdrawn as long as it is identifiable. Your decision about whether or not to participate or to stop participating will not jeopardize your future relations with Auburn University, the Department of Kinesiology, or the Sport Biomechanics Laboratory.

Your privacy will be protected. Any information obtained in connection with this study will remain confidential. Information obtained through your participation may be published in a professional journal or presented at a professional meeting.

If you have questions about this study, *please ask them now or* contact Andrea Sumner at sumneam@auburn.edu, or Dr. Wendi Weimar at weimawh@auburn.edu. Furthermore, you may reach any of these three individuals by calling the Sport Biomechanics Laboratory at 844-1468. A copy of this document will be given to you to keep.

If you have questions about your rights as a research participant, you may contact the Auburn University Office of Human Subjects Research or the Institutional Review Board by phone (334)-844-5966 or e-mail at hsubjec@auburn.edu or IRBChair@auburn.edu.

HAVING READ THE INFORMATION PROVIDED, YOU MUST DECIDE WHETHER OR NOT YOU WISH TO PARTICIPATE IN THIS RESEARCH STUDY. YOUR SIGNATURE INDICATES YOUR WILLINGNESS TO PARTICIPATE.

______________________________	______________________________
Participant's signature Date	Investigator obtaining consent Date
______________________________	______________________________
Printed Name	Printed Name

Appendix C

ANOVA Table for the Effect of Load Carriage on Joint Kinematics

Stationary		**Mean**	**SD**	***p***	***η2***	***Power***
TA_{AVG} (°)						
	U	18.70	3.39			
	SDS	21.82	4.55	< 0.001	0.499	0.996
	MDS	21.94	4.33			
HA_{AVG} (°)						
	U	73.82	4.06			
	SDS	77.08	4.14	0.020	0.228	0.719
	MDS	78.09	6.38			
AA_{AVG} (°)						
	U	91.35	3.09			
	SDS	93.75	3.24	< 0.001	0.684	1.000
	MDS	93.76	3.34			

Dynamic		**Mean**	**SD**	***p***	***η2***	***Power***
TA_{AVG} (°)						
	U	15.79	3.73			
	SDS	22.96	5.43	< 0.001	0.855	1.000
	MDS	22.71	5.10			
HA_{AVG} (°)						
	U	86.48	2.49			
	SDS	88.30	3.16	0.018	0.236	0.739
	MDS	88.72	3.49			
AA_{AVG} (°)						
	U	90.35	3.61			
	SDS	90.51	3.06	0.475	0.048	0.167
	MDS	89.80	3.29			

Appendix D

ANOVA Table for the Effect of Load Carriage on Joint Kinematics

Foot Flat		**Mean**	**SD**
KA_{AFF} (°)			
	U Pre	176.45	1.76
	SDS Pre	175.80	2.80
	MDS Pre	175.17	2.94
KA_{AFF} (°)	U Post	176.21	2.67
	SDS Post	175.54	2.62
	MDS Post	175.47	2.43

ANOVA for Loading Condition: ($F(2,14) = 1.114$, $p = 0.341$, $\eta 2 = 0.069$, Power = 0.227)

ANOVA for Time: ($F(2,14) = 0.125$, $p = 0.729$, $\eta 2 = 0.008$, Power = 0.063)

Foot Flat		**Mean**	**SD**
AA_{AFF} (°)			
	U Pre	87.92	3.81
	SDS Pre	87.51	2.90
	MDS Pre	87.53	2.81
AA_{AFF} (°)	U Post	88.48	3.98
	SDS Post	88.56	4.21
	MDS Post	87.40	2.26

ANOVA for Loading Condition: ($F(2,14) = 0.467$, $p = 0.631$, $\eta 2 = 0.030$, Power = 0.119)

ANOVA for Time: ($F(2,14) = 0.691$, $p = 0.419$, $\eta 2 = 0.044$, Power = 0.122)

Appendix E

ANOVA Table for the Effect of Load Carriage on Average ES sEMG

Pre		**Mean**	**SD**	***p***	***η2***	***Power***
EMG_{AVG} (%MVIC)						
	U	1.70	1.11			
	SDS	3.57	1.73	< 0.001	0.674	1.000
	MDS	3.64	1.67			
Post		**Mean**	**SD**	***p***	***η2***	***Power***
EMG_{AVG} (%MVIC)						
	U	1.72	1.03			
	SDS	4.01	1.92	< 0.001	0.710	1.000
	MDS	3.69	1.65			

ANOVA Table for the Effect of Load Carriage on Peak ES sEMG

Stationary		**Mean**	**SD**	***p***	***η2***	***Power***
EMG_{PEAK} (%MVIC)						
	U	3.05	1.17			
	SDS	5.61	2.33	< 0.001	0.665	1.000
	MDS	5.30	2.52			
Dynamic		**Mean**	**SD**	***p***	***η2***	***Power***
EMG_{PEAK} (%MVIC)						
	U	6.62	3.60			
	SDS	9.74	5.37	< 0.001	0.654	1.000
	MDS	10.25	5.25			

Appendix F

ANOVA Table for the Effect of Load Carriage on Peak Contact Pressure

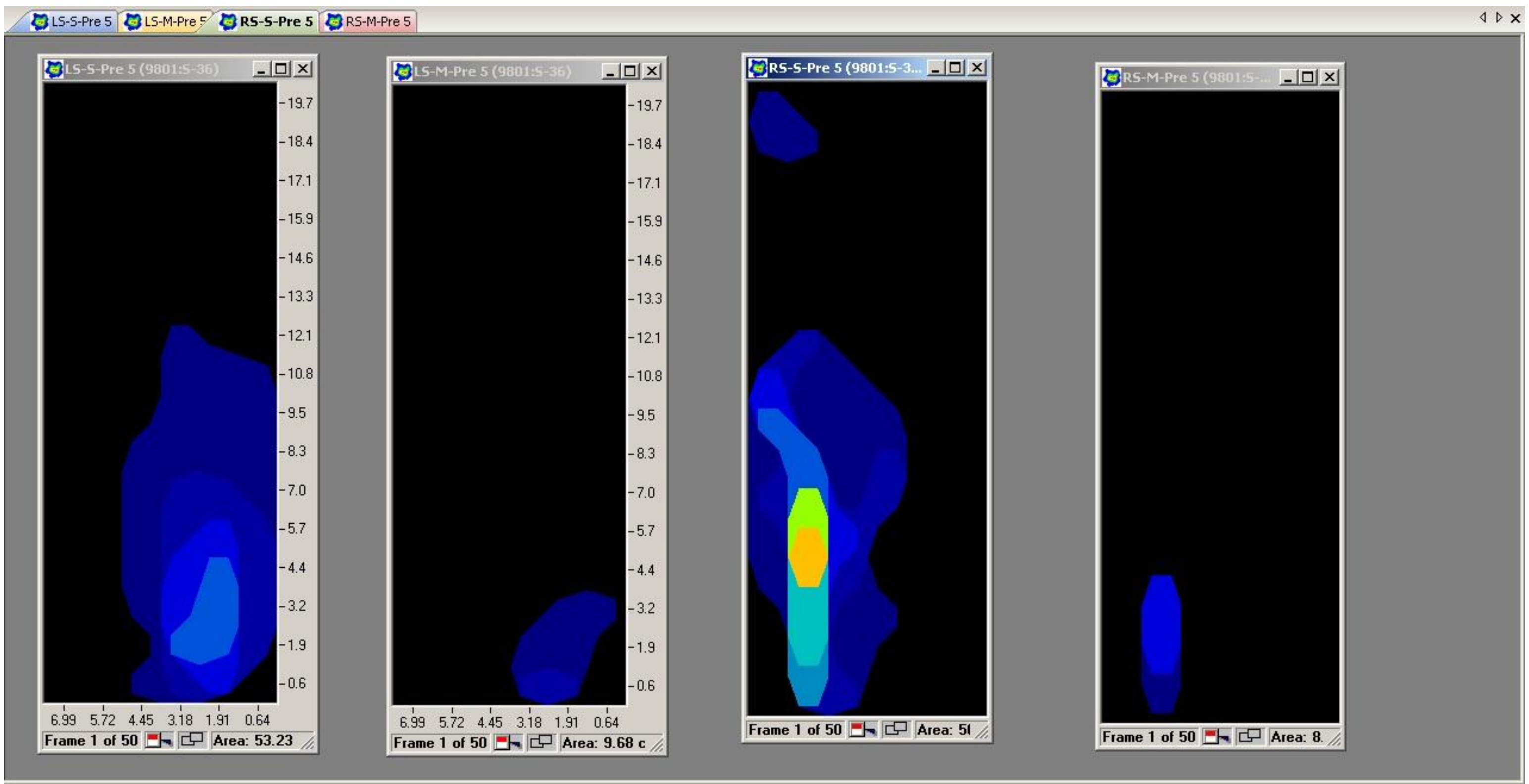
		Mean	**SD**
$SHLDR_{PEAK}$ (kPa)			
	SDS	11.59	5.80
	MDS	8.89	5.16
ABS_{PEAK} (kPa)	SDS	4.77	1.30
	MDS	5.48	1.47

ANOVA for Body Location x Loading Condition :

($F(1,15) = 8.993$, $p = 0.009$, $\eta^2 = 0.375$, *Power* = 0.800)

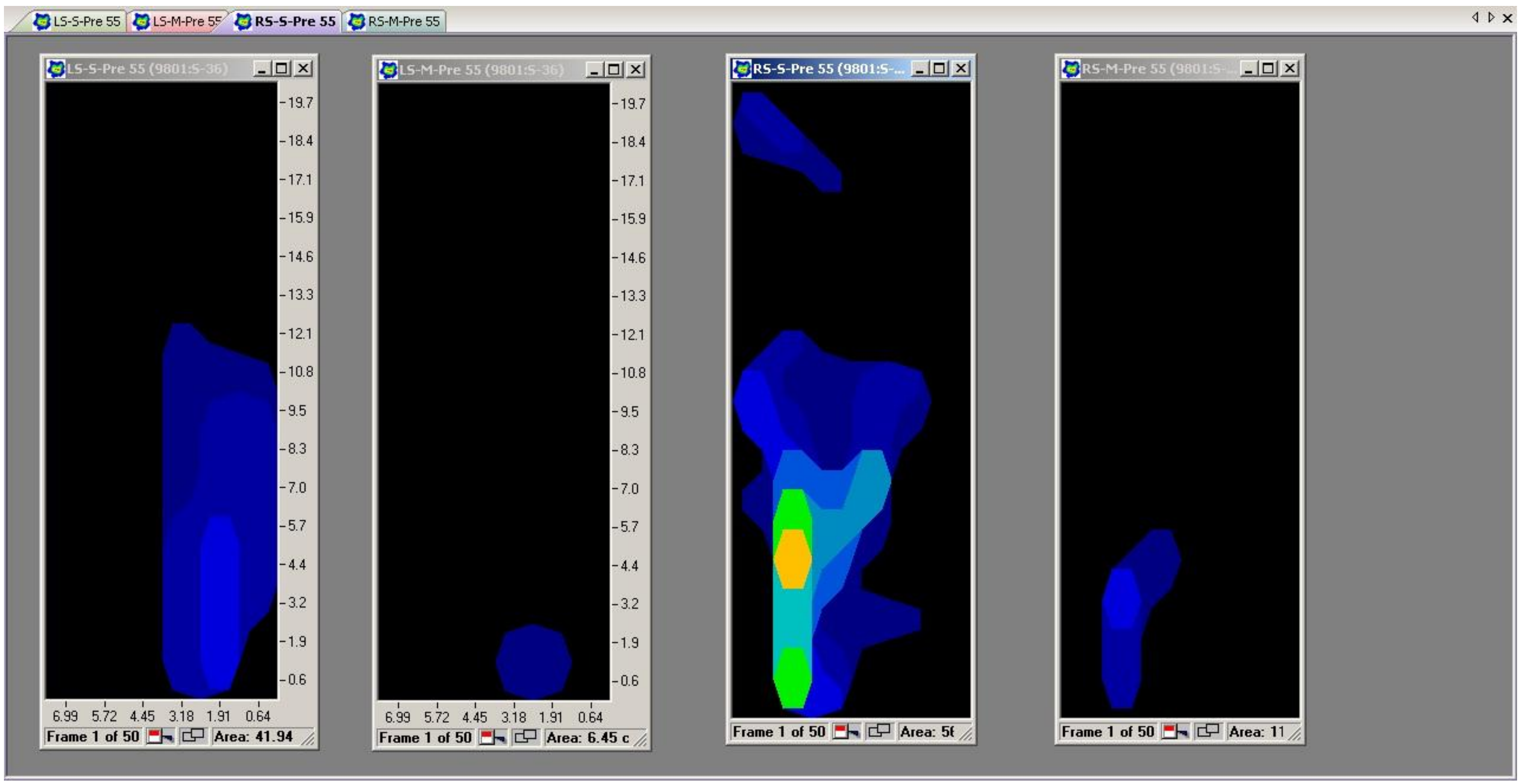
Appendix G

Contact Pressure Output at the Left and Right Shoulders for the SDS (LS-S-Pre5) & MDS (LS-M-Pre5) Loading Conditions during the Pre5 Measurement



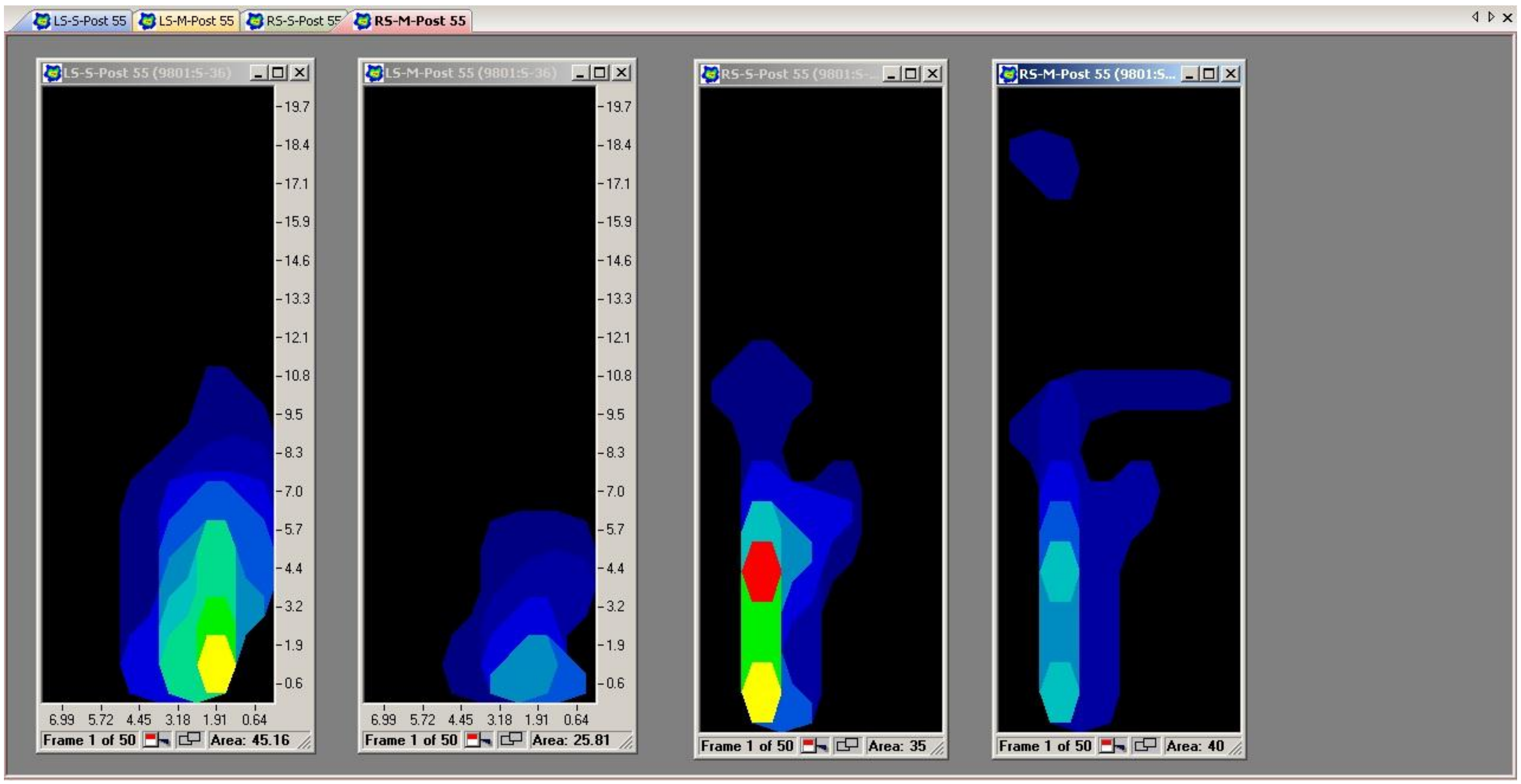
Appendix H

Contact Pressure Output at the Left and Right Shoulders for the SDS (LS-S-Pre55) & MDS (LS-M-Pre55) Loading Conditions during the Pre55 Measurement



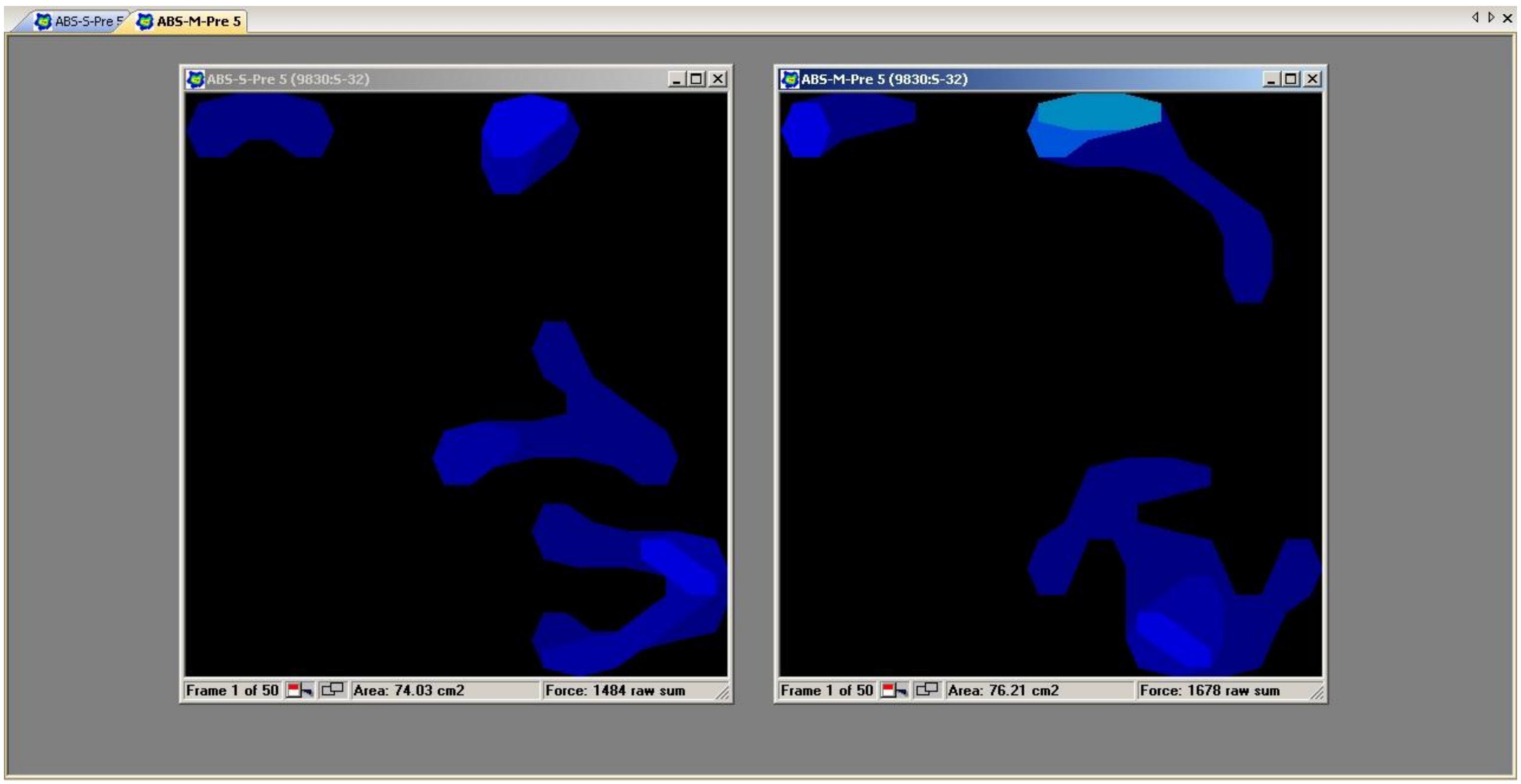
Appendix I

Contact Pressure Output at the Left and Right Shoulders for the SDS (LS-S-Post55) & MDS (LS-M-Post55) Loading Conditions during the Post55 Measurement



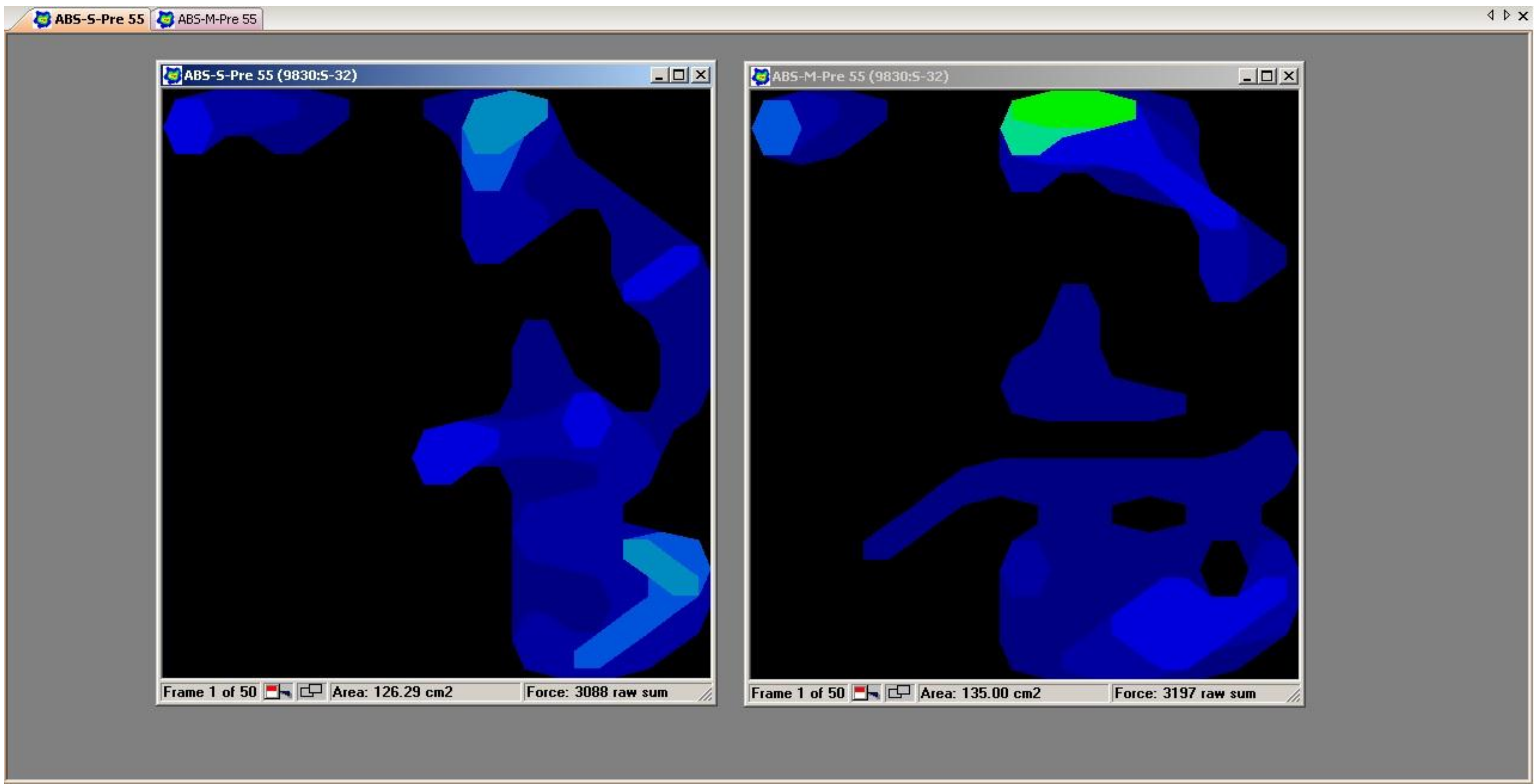
Appendix J

Contact Pressure Output at the Abdominal Region for the SDS (ABS-S-Pre5) & MDS (ABS-M-Pre5) Loading Conditions during the Pre5 Measurement



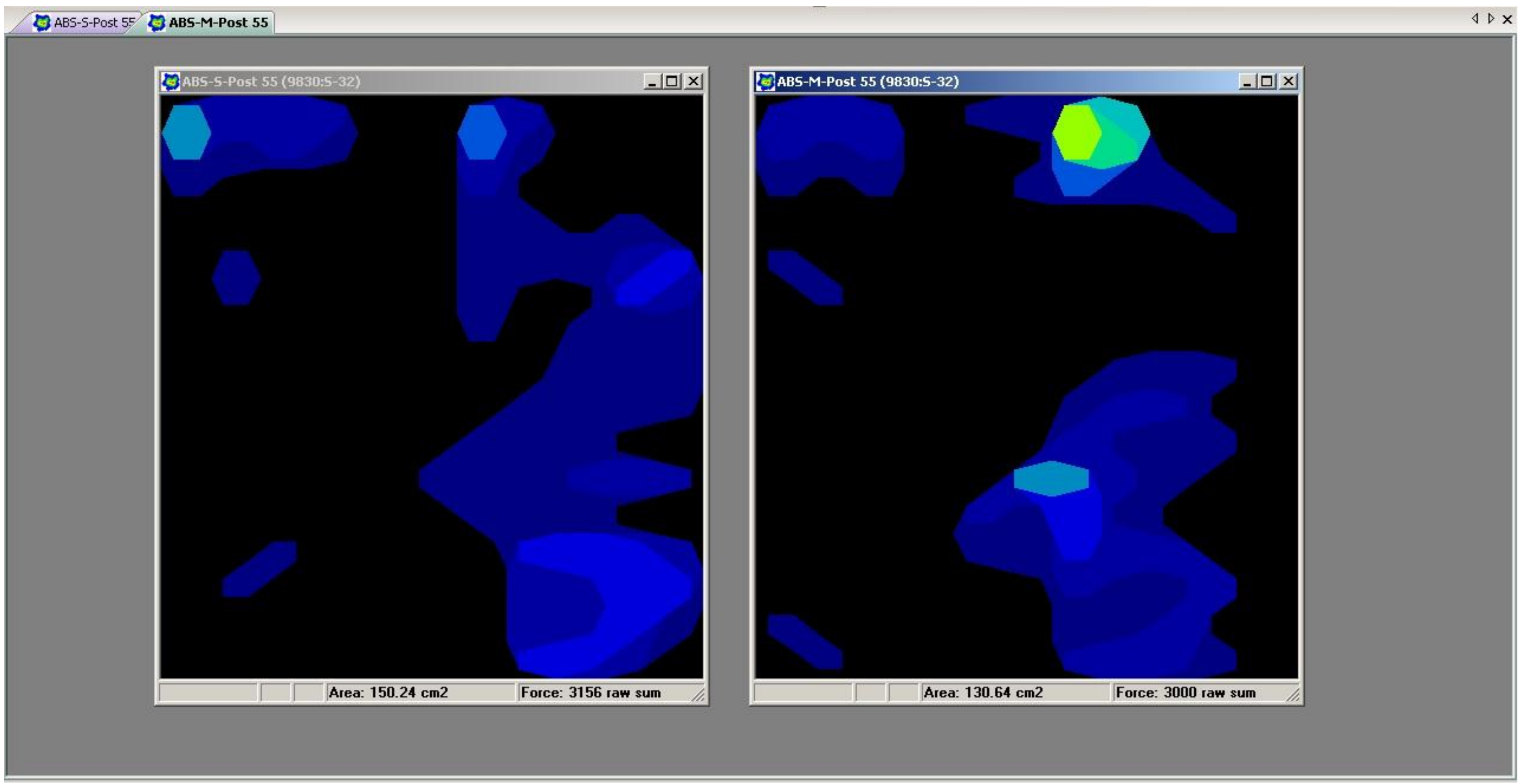
Appendix K

Contact Pressure Output at the Abdominal Region for the SDS (ABS-S-Pre55) & MDS (ABS-M-Pre55) Loading Conditions during the Pre55 Measurement



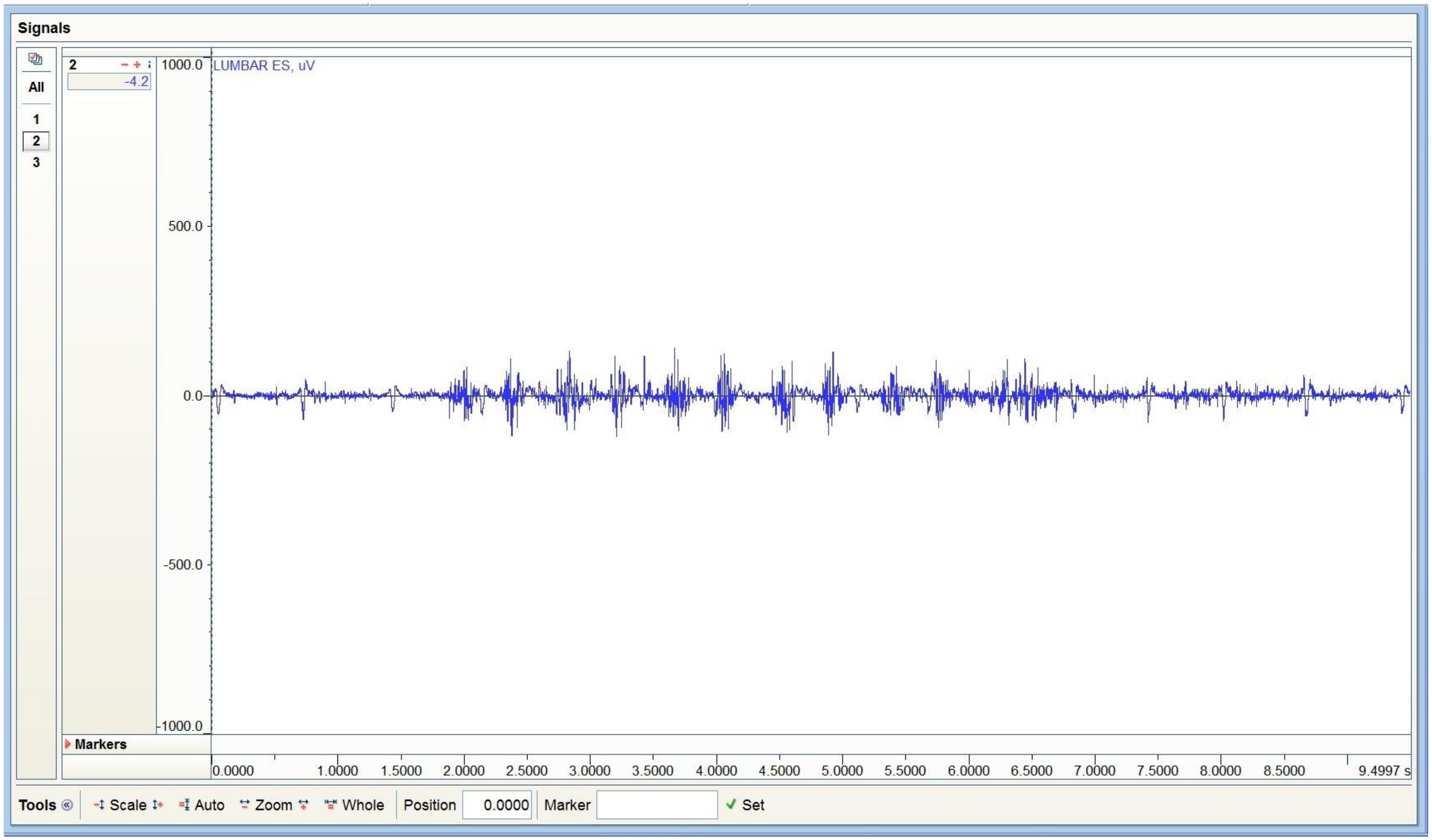
Appendix L

Contact Pressure Output at the Abdominal Region for the SDS (ABS-S-Post55) & MDS (ABS-M-Post55) Loading Conditions during the Post55 Measurement



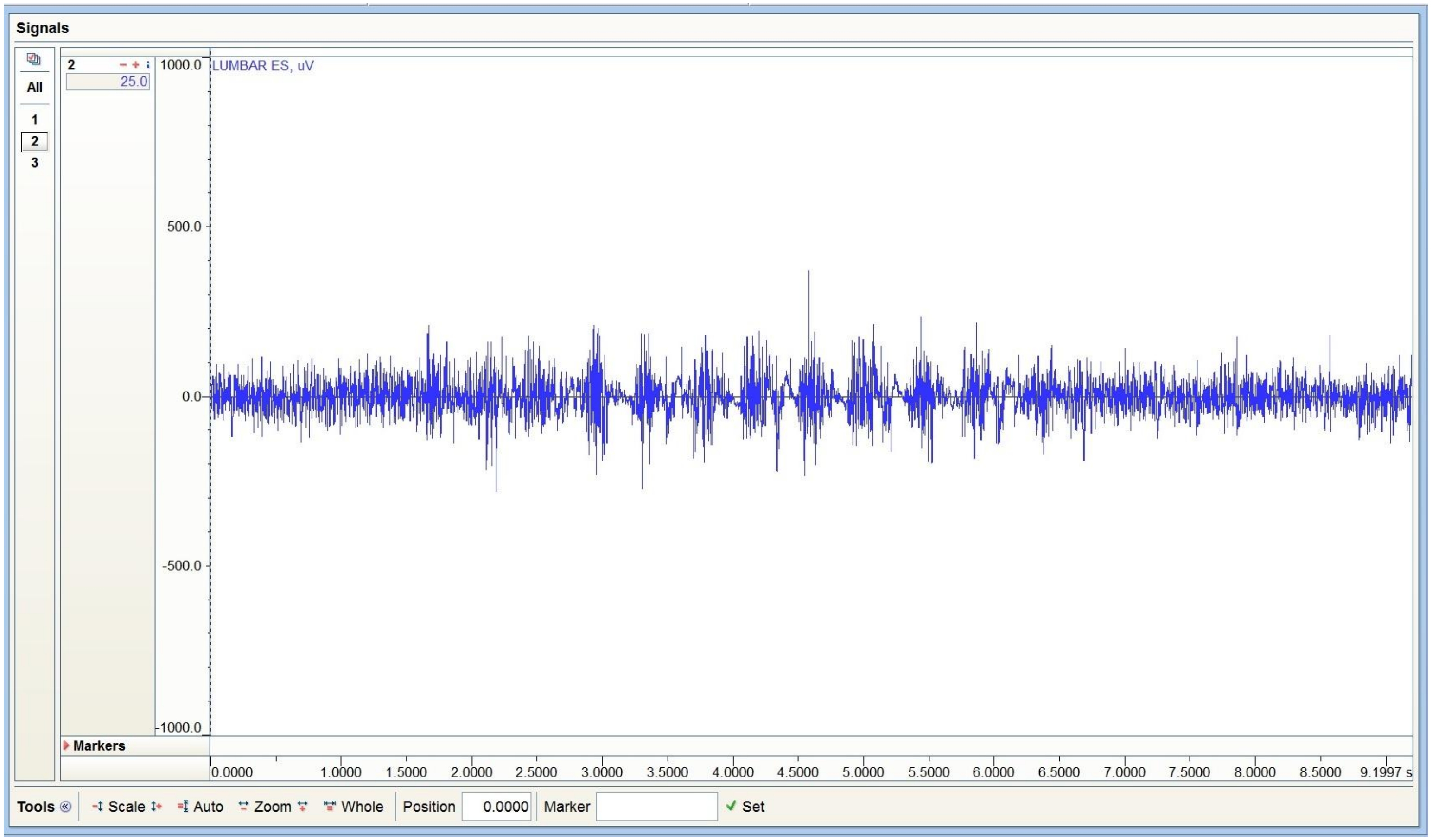
Appendix M

Raw sEMG Signal of Erector Spinae for Whole Gait Trial during the U Loading Condition



Appendix N

Raw sEMG Signal of Erector Spinae for Whole Gait Trial during the SDS Loading Condition



Appendix O

Raw sEMG Signal of Erector Spinae for Whole Gait Trial during the MDS Loading Condition