THESIS

LOWER EXTREMITY KINEMATICS DURING WALKING WITH SNOWSHOES

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WE HEREBY RECOMMEND THAT THE THESIS PREPARED UNDER OUR SUPERVISION BY REBECCA N. KURTZ ENTITLED LOWER EXTREMITY KINEMATICS DURING WALKING WITH SNOWSHOES BE ACCEPTED AS FULFILLING IN PART REQUIREMENTS FOR THE DEGREE OF MASTER OF SCIENCE.

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ABSTRACT OF THESIS

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The popularity of snowshoeing as a form of winter recreation has increased in recent years, due, in part to the development of lightweight snowshoes that provide flotation, traction and stability. However, there is little data on the effects of snowshoes and their design on the biomechanics of walking. **PURPOSE**: To determine the effects of snowshoes and their frame design on lower extremity kinematics during level and downhill walking. We hypothesized that: 1) lower extremity sagittal plane kinematics would be altered during level snowshoe vs. overground walking; 2) during downhill walking, the use of a snowshoe with a flexible tail frame (flex tail) would reduce ankle plantarflexion and knee flexion angular velocities during early stance. **METHODS**: Twelve adults (6 males, 6 females, age, 26.7±5.6 yrs, body mass = 68.5±10.7 kg) with prior snowshoe experience completed six, 3-minute level walking trials. Three of the trials were at a walking speed of 1.4 m/s and three were at 0.9 m/s. Subjects walked on packed snow using conventional snowshoes and flex tail snowshoes and overground without snowshoes. In addition, each subject completed two downhill (14º grade) walking trials at a self-selected speed using conventional and flex tail snowshoes. We placed lightweight inertial/gyroscopic sensors on the sacrum, posterior mid-thigh, posterior mid-leg and on the dorsal aspect of the foot. During each trial, we recorded sensor orientation and calculated hip, knee and ankle joint angles and angular velocities.
RESULTS: Participants had greater hip and knee flexion during stance and greater hip flexion during swing while snowshoe vs. overground walking on level ground at 1.4 m/s. At 0.9 m/s, subjects had a greater amount of knee flexion during snowshoeing. In addition, ankle plantarflexion began earlier in the gait cycle during snowshoe vs. overground walking. Lower extremity kinematics were similar across snowshoe frame designs during level and downhill walking. **CONCLUSIONS**: Our results suggest that

snowshoeing on packed snow alters lower extremity kinematics resulting in a more flexed leg compared to overground walking. The gait kinematics adopted during snowshoeing may reflect a strategy to limit the effects of having an extended heel on ankle plantarflexion during heel strike.

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TABLE OF CONTENTS

PREFACE

This document contains a manuscript embedded in a thesis. The first three chapters contain the thesis introduction, literature review, and methods. The manuscript follows in the format of the journal to which it was submitted. The following pages consist of figure legends, a table, and figures. Appendix B with the informed consent form used for subjects is the final section of this thesis.

CHAPTER I

INTRODUCTION

Snowshoeing is a popular winter activity, both as a competitive and recreational sport. The number of individuals who snowshoe continues to increase, reaching ~5 million participants in 2006 (Outdoor Industry Foundation, 2006). Snowshoeing's popularity is due, in part, because it is an easy-to-learn, low-impact recreational activity. In addition, improvements in design have resulted in lightweight snowshoes that are easy to use because they provide adequate traction, flotation, and stability. The metabolic rate during snowshoeing is ~2x greater than during overground walking, making it an excellent form of exercise (Hall, et al., 2004; Schneider, et al., 2001). The increase in metabolic rate during snowshoeing is likely caused by a combination of altered biomechanics, added mass of the snowshoe and walking on a soft surface. However, the biomechanics of walking with snowshoes have not been reported. Quantifying lower extremity biomechanics during snowshoe walking may help explain the greater metabolic rate during snowshoeing, provide insights into locomotion motor control strategies, and facilitate the development of appropriate training methods to prepare for snowshoeing. In addition, evaluating how snowshoe frame design affects lower extremity biomechanics may provide guidance for further product improvements.

Lower extremity kinematics have not been described during snowshoeing. As a result, it is not clear the degree to which motor control strategies must be adjusted when walking on snowshoes. However, studies that have examined kinematics during level walking combined with body-weight support, added foot mass, or a soft surface, provide some insights into potential kinematic adjustments. The position and trajectory of the foot is tightly controlled during normal (Winter, 1992) and body-weight supported walking (Ivanenko, et al., 2002), most likely to minimize metabolic cost and to prevent tripping and slipping. The addition of a snowshoe is likely to require adjustments in foot trajectory and therefore lower extremity kinematics. Adding mass to the foot during level, treadmill walking increased stride length and swing time and altered segment elevation angles at toe-off (Browning, et al., 2007). Lejeune, Willems, & Heglund (1998) reported the mechanics of walking in loose sand vs. overground. Although Lejeune et al. (1998) did not report kinematics, the differences in center of mass gravitational potential energy between walking in sand and overground suggest changes in lower extremity joint angles (Lejeune, et al., 1998).

It seems reasonable that the addition of a snowshoe to the foot would alter lower extremity kinematics during walking. Most snowshoes are attached to the foot via a binding that allows the snowshoe to rotate around the attachment point, just inferior to the metatarsals. As a result, the snowshoe may drag along the snow as the foot swings forward. While this design allows accumulating snow to slide off the back of the snowshoe during leg swing, the extended, rigid tail of the snowshoe may alter lower extremity kinematics, particularly during swing and early stance. As the foot swings forward and the ankle dorsiflexes, the heel typically contacts the deck (top surface) of the snowshoe, which may force the foot into plantarflexion, essentially beginning heel strike earlier in leg swing. Snowshoers may accommodate for the extended heel by lifting the foot higher, using a more flexed hip and knee throughout swing, and keeping the ankle more plantarflexed prior to heel strike. Frontal plane kinematics are also likely to be different during snowshoe vs. overground walking, because of the additional width of the snowshoe frame. The wider snowshoes would be expected to increase step width, especially during slower walking.

The kinematic adjustments caused by walking in snowshoes may become more pronounced when walking downhill, as the extended heel would cause rapid ankle plantarflexion once the heel contacts the snowshoe deck. Downhill walking results in a more flexed posture at the hip, knee and ankle compared to overground level walking. The ankle and knee angular velocities are also greater during early stance in downhill vs. level walking (Redfern & DiPasquale, 1997). A snowshoe design that incorporates a flexible tail may reduce the need to alter kinematics, particularly during downhill walking, as the tail would flex during late swing and early stance. Presumably, a kinematic pattern more representative of overground downhill walking would be more comfortable for the participant.

Statement of Problem

The purpose of this study was to determine the effects of snowshoes and their frame design on lower extremity kinematics during level and downhill walking.

Hypotheses

We hypothesized the following: 1) lower extremity sagittal plane kinematics would be different during level snowshoe vs. overground walking and 2) during downhill walking; the use of a snowshoe with a flexible tail frame would reduce ankle plantarflexion and knee flexion angular velocities during early stance.

CHAPTER II

LITERATURE REVIEW

Participation in snow sports has increased in recent decades. Popular forms of winter recreation include downhill skiing, snowboarding, cross country skiing, and snowshoeing. The number of individuals who snowshoe during a year has increased to approximately 5 million in North America (Outdoor Industry Foundation, 2006). This increase is likely due to a change in the structure of snowshoes. Lightweight snowshoes have evolved that offer greater traction, flotation, and stability. Smaller aluminum and plastic frames as well as efficient binding systems provide for ease of locomotion. As a result, snowshoeing is a popular form of winter exercise. Further improvements in snowshoes may allow for greater accessibility to a larger number of people. However, no data currently exists concerning the kinematics of walking in snowshoes. To further improve design and provide insights for performance and injury prevention, it is necessary to understand the gait pattern while walking with snowshoes.

Overground Walking Kinematics

During overground walking, the stance phase is approximately 62% of the gait cycle. The swing phase is the remaining 38% of the gait cycle. An increase in the velocity of walking decreases each gait cycle duration (time). In addition, an increase in velocity also leads to a subsequent increase in stride frequency and stride length. As walking speed increases, the first component to have a large increase is stride length. The length of a stride can only increase to a point, after this point; to further increase speed stride frequency (cadence) must become greater. The percent of the gait cycle spent in stance also decreases significantly as walking speed increases. Similarly, the percent of the gait cycle spent in swing increases. Double limb support time decreases as speed increases, because of the fact that the body is moving towards a flight phase observed in running (Murray, Mollinger, Gardner, & Sepic, 1984; Kirtley, Whittle, & Jefferson, 1985). Murray, et al., (1985) show how the gait cycle changes based on the velocity in normal healthy females: at a velocity of 0.83 m/s, step length is approximately 0.557 m and cadence is 87 steps per minute. At a speed of 1.42 m/s, step length is .703 m and cadence is 117 steps per minute.

There are eight sub-phases of gait: initial contact, loading response, mid stance, terminal stance, pre-swing, initial swing, mid swing, and terminal swing. During initial contact, the leg is positioned to begin stance. Initial contact is also the first period of double stance in the gait cycle. The loading response involves shock absorption due to weight transfer from one limb to the other. During mid stance, progression of the limb over the foot occurs via ankle dorsiflexion in combination with knee and hip extension. Terminal stance begins with heel rise. Weight continues to move ahead to the forefoot during this phase. Pre-swing involves the second period of double stance and weight transfer to the opposite foot. Initial swing begins with the lifting of the foot from the floor and continues with the swinging of the foot forward. Mid swing occurs when the swing foot is opposite the stance foot and ends when the tibia is vertical to the ground. When the limb advancement is completed by full knee extension, terminal stance has occurred (Perry, 1992, Rose & Gamble, 1994; Winter; 2009).

During normal gait, the gait cycle begins with heel strike. At heel strike, the foot is typically in a neutral position or slightly plantarflexed. Rapid plantarflexion occurs immediately post heel strike, and at foot flat, dorsiflexion begins. During late stance, plantarflexion begins in preparation for toe off. After toe off, dorsiflexion occurs and a neutral foot position is achieved by mid swing. The knee is almost fully extended at heel strike and begins flexion at heel strike. In midstance, the knee begins to extend as the center of mass moves over the foot. At terminal stance, the knee undergoes flexion again and continues through mid swing. During late swing, the knee extends again in preparation for heel strike. The hip is flexed at heel strike and progressively begins to extend during mid stance. At toe off and the onset of swing, the hip begins to flex and continues through flexion until the next heel strike (Perry, 1992; Murray, Drought, & Kory, 1964).

In walking, the range of motion of the ankle joint (articulation between the tibia and talus) is approximately 30° (Alton, Baldey, Caplan, & Morrissey, 1998). In the first 10% of stance, the ankle plantarflexes 5-10° as the forefoot descends to the ground post-heel strike. Once foot flat occurs, the ankle moves into dorsiflexion as the tibia moves over the ankle joint. During stance, the ankle dorsiflexes approximately 10°, which is the maximum amount of dorsiflexion observed during the gait cycle. At about 50% of the gait cycle, heel rise occurs and the ankle moves into plantarflexion. The maximum amount of plantarflexion, 30°, is observed at the end of stance. The final dorsiflexion phase occurs in the first portion of swing. Finally, during the last half of swing the foot remains in a neutral position (Perry, 1992). Additionally, during slow walking, the amount of dorsiflexion achieved during stance increases significantly. Plantarflexion is greater during late stance for fast walking compared to slower walking (Murray, Mollinger, Gardner, & Sepic, 1984).

Knee range of motion during walking is approximately 70°. At heel strike, the knee is typically flexed about 5° and subsequently flexes with loading approximately 20°. The knee then extends as stance proceeds. Flexion begins again at 40% of the gait cycle and continues to flex at toe off and throughout the first half of swing. Maximum flexion during swing is approximately 60°. Knee extension resumes in mid swing in preparation for heel strike. Maximum knee extension is attained prior to heel strike as the knee then slightly flexes just before heel strike (Perry, 1992). During slower walking, the knee is significantly more extended at heel strike than at faster walking speeds. The knee also does not undergo as much flexion during stance during slow walking. The increased knee flexion during faster walking is likely due to a shock-absorbing mechanism from a faster onset of weight loading on the stance leg (Murray et al., 1984).

The hip is typically in maximum flexion at heel strike. After foot contact, the hip extends as the center of mass moves over the foot. Typical hip range of motion is approximately 40°. At heel strike, the hip is flexed to about 30° and undergoes extension throughout stance as the pelvis rotates over the support leg. Maximum extension of 10° is attained at 50% of the gait cycle. Flexion follows this period of extension and continues into swing, where it reaches approximately 35° of flexion (Perry, 1992). During slower walking, the hip is less flexed at heel strike and subsequently extends less at toe-off (Murray et al., 1984).

The inverted pendulum theory proposes that the center of mass travels in an arc trajectory. The stance leg, during single support, acts like an inverted pendulum to move the center of mass forward. This inverted pendulum serves to conserve energy during walking via the exchange of kinetic and potential energy. Less energy is expended when the stance leg performs similar to a pendulum. The stance leg does not act like a passive pendulum, rather, energy is required to accelerate and decelerate the leg (Kuo, 2007; Cavagna, Heglund, & Taylor, 1977). The inverted pendulum theory states that there are energetic advantages to walking with a more stiff leg as opposed to a significantly more bent leg that minimizes the vertical travel of the center of mass (Lee & Farley, 1998).

Minimizing energetic cost during walking involves the transfer of kinetic and gravitational potential energies. However, the exchange of energy is not a complete exchange. Thus, external work must be done to continue walking at the same speed. Potential energy reaches its peak value during midstance when the center of mass rises (Cavagna & Margaria, 1966; Winter, 1979). Between heel strike and midstance, kinetic energy is transformed to potential energy as the forward velocity of the center of mass decreases. Between midstance and toe off, gravitational energy is converted to kinetic energy as the forward velocity of the center of mass increases (Lee & Farley, 1998). This energy is either absorbed or produced by the various segments at different periods during the gait cycle. On a hard surface, this energy exchange is very efficient. A large portion of the potential energy is converted to kinetic energy to allow for propulsion of the limb to advance the body. The primary need for energy occurs at toe off to propel the limb into swing and achieve ground clearance to avoid tripping. The bulk of the power generated for toe off comes from ankle plantarflexion near the end of stance. Deceleration of the leg during swing comes from energy absorption by the hip extensors and knee flexors. Weight acceptance at heel strike also involves a period of energy absorption (Winter & Robertson, 1978).

When walking, weight is transferred between the two legs. Weight moves forward on the stance leg when the body is allowed to fall forward. Lifting the swing foot allows the body to utilize the change in posture for body propulsion. At the end of the single support period, the swing limb is significantly in front of the body's center of mass. This creates a relatively unstable situation where the body is falling to the floor with the swing limb. The body descends to the floor at a rapid rate, thus, shock absorption in the swing limb is important. Plantarflexion immediately after heel strike helps to decrease the rate at which the foot impacts the floor. Knee flexion in early stance also aids in shock absorption. The quadriceps contract to prevent further knee flexion and in the process a portion of the loading force is shifted to the quadriceps (Winter & Robertson, 1978).

Motor Control during Walking

Kinematic patterns during walking are a result of motor commands and adjustments in kinematics associated with walking with snowshoes should reflect changes in motor control strategies. During walking, the challenge is to move the body outside of the base of support from standing while avoiding a fall. In steady speed walking, the center of mass is frequently outside of the base of support. During quiet standing, balance is achieved via the ankle plantarflexors and dorsiflexors in the anterior-posterior direction as well as the hip abductor and adductor muscles in the medio-lateral direction. Yet, walking provides a different challenge to the central nervous system. To initiate walking, individuals need to voluntarily accelerate the center of gravity by falling forward. To terminate walking, we must voluntarily brake to return the center of gravity to the base of support. The body must also protect against vertical collapse during walking. To collapse, the hip, knee, and ankle must undergo flexion. Hence, the extensor muscles in the leg are the body's preventative mechanism against collapse. This support moment is the combination of joint moments at the hip, knee, and ankle. Winter (1995) observed a high amount of variability in the knee and hip moments across trials of the same subjects. However, as the hip moment changed, the knee and ankle moments were changed so that the overall support moment remained relatively consistent across trials.

McGeer (1993) describes human gait as "passive dynamic walking" in which gait can be sustained by the interaction of gravity and inertia. Motor control occurs during initiation and termination of gait, as well as when there is a need to alter gait due to uneven terrain or for an energy source. Three major sensory systems, visual, vestibular and somatosensory, are involved in the control of walking. Vision helps to plan locomotion. The vestibular system senses changes in linear and angular accelerations of the head, while the somatosensory system senses position and velocity of body segments, as well as their impact with the ground and other external objects (Winter, 1995). Muscle spindles within the skeletal muscle are also essential in motor control of gait. These muscle spindles have the ability to detect stretch in a muscle, which then subsequently tells the muscle to contract to avoid overstretching and injury. In addition, they provide continuous feedback on muscle length and changes in muscle length (Winter, 1995). During walking, the ankle muscles do not play a large role in balance, as they cannot prevent a fall. The ankle muscles can only slightly alter the anterior-posterior and medio-lateral acceleration of the center of gravity- (Winter, 1995).

Royer and Martin (2001) looked at the effects of various leg loads and their effects on motor control strategies during the swing phase of walking. They determined that walking with leg loads results in changes in motor control strategies. The leg with the additional mass experiences a greater swing time than the leg without the additional mass. This seems to suggest that motor control strategies do not adapt to additional loads.

Muscle Mechanical Properties

Force-Length

The ability to produce force largely depends on the length of the muscle at time of initiation of force production as well as the velocity at which the muscle contraction occurs. If there is a pre-stretch of the muscle just prior to a concentric contraction, the number of cross-bridges that can occur throughout the entire contraction is greater, resulting in greater force production (Rassier, MacIntosh, & Herzog, 1999). The level of isometric tension that can be produced depends on the length of the muscle at the time of force production (Wilkie, 1950). The length dependence of muscular contractions relates to the sliding filament and cross-bridge theories. The sliding filament theory is based on the fact that actin and myosin slide along one another to create changes in length of the sarcomere, which results in changes in length of the muscle. The cross-bridge theory suggests that the sliding of actin and myosin along one another is caused by cross bridges capable of producing force. Therefore, the more cross bridges that are engaged, the greater the force produced. Thus, the point at which there is maximum overlap between the actin and myosin filaments should be the point at which the greatest force is produced by the sarcomere. Anything beyond this overlap results in a decrease in the number of cross-bridges, which results in a decrease in force production. When the muscle length is shorter than the length of optimal overlap of the sarcomeres, force production decreases. Greater muscle activation occurs with a greater amount of troponin bound to calcium (Rassier, MacIntosh, & Herzog, 1999).

Muscles function on the ascending or descending limb of the force-length curve and reach their plateau in force output at approximately the mid range of motion of the joint. In a stretch-shortening cycle, the muscle is initially stretched as it increases in force and then subsequently shortens as force output decreases. In a shortening-stretch cycle, the muscle shortens as force is created and then stretches as forces declines. It has been observed that muscles going through a stretch-shortening cycle operate on the descending portion of the force-length curve, while muscles undergoing a shortening-stretch cycle exist on the ascending portion of the force-length curve. Additionally, muscles that ~~are~~ function on the ascending portion of the force-length curve tend to have passive force during short muscle lengths, while muscles that function on the descending portion generally have passive forces that appear at long muscle lengths. The surrounding connective tissues have the ability to stretch, which allows passive force in the muscle to occur. The number of sarcomeres in a muscle fiber also affects the amount of force that can be produced over a range of motion. Specifically, downhill training (walking or running) seems to increase the number of sarcomeres in the quadriceps muscles (Rassier, MacIntosh, & Herzog, 1999).

Force-Velocity

Force output depends on the velocity of the muscle contraction. A slower velocity results in more cross-bridges attaching, which increases the force of contraction. As such, faster rates of muscle contraction result in fewer cross-bridges, which decreases the amount of force production. Abbott & Wilkie (1953) have shown that during a concentric muscle contraction, as velocity of shortening increases, the amount of force the muscle is able to produce decreases. As the opposing force decreases, the muscle can be shortened at a faster rate (Wilkie, 1950). This decrease in force production is not linear, but rather it is an exponential decrease that occurs with an increase in velocity of contraction (Fenn & Marsh, 1935). Maximum force occurs when the muscle can no longer be shortened and the velocity is zero, such as in an isometric contraction (Wilkie, 1950). However, this is not true for an eccentric muscle contraction. During an eccentric muscle contraction, as velocity of lengthening increases, force increases as well (Abbott & Wilkie, 1953). Additionally, the average amount of torque at the knee produced by an eccentric contraction is significantly greater than the amount of torque that is able to be produced by a concentric contraction (Westing, Cresswell, & Thorstensson, 1991). However, EMG activity has been observed to be lower during eccentric action in the vastus medialis, vastus lateralis, and rectus femoris compared to that of concentric muscle action at the same velocity. This discrepancy between EMG activity levels and torque produced during eccentric contractions is hypothesized to be due to reduced neural drive. If the force output is similar, there is not as much muscle activation necessary to produce the same force eccentrically as is needed concentrically (Westing, Cresswell, & Thorstensson, 1991). In addition, it has been shown that oxygen consumption increases with higher velocities of shortening at a constant force; however, there is no significant difference in oxygen consumption when the muscles are stretched during contraction with increasing velocities. When the velocity of a muscle contraction is kept constant, a linear relationship between electrical activity of the muscle and the amount of weight lifted exists during both shortening and lengthening of the muscle. However, the slope of the line is less during lengthening than that of shortening (Bigland & Lippold, 1954).

Methodology to Quantify Kinematics

Inertial measurement units (IMU) are a novel method of collecting kinematic data. These are an alternative to using video-based motion capture. Inertial measurement units may be an easier method of collecting data in the field as more data can be collected continuously during each trial with the inertial measurement units. A video camera has to remain stationary during data collection, with the exception of pan and tilt systems that allow a larger field of view to be covered. Thus, it is only possible to collect a few steps of data each time the subject is within view. Additionally, the ability to accurately and easily collect data in the field allows for a greater variety of settings and terrain that can used to assess kinematics. IMU's also eliminate several limitations presented with a laboratory motion capture system. In a laboratory, subjects are confined to a treadmill or limited amount of space. Thus, movement analysis over a variety of terrains is restricted. This can reduce the application of some research to real-life settings. It has been established that data obtained from an inertial measurement unit correlates well (no significant differences) with an optical motion capture system of which the accuracy is well recognized (Pfau, Witte, & Wilson, 2005). However, movements at higher speeds, such as jogging or running, resulted in a larger amount of offset from the inertial measurement unit (Pfau, Witte, & Wilson, 2005).

IMU's have a variety of applications, including assessing activities of daily living, mechanical work, kinematic assessment, and neurological disorders. IMU's contain a three dimensional accelerometer, three dimensional rate gyroscope, and three dimensional magnetometer to measure linear acceleration ,~~and~~ angular velocity, and compass heading. ~~Use of~~With a tri-axial accelerometer ~~allows for~~ direct measurement of acceleration is made in three dimensions. The accelerometer can be used to determine~~s~~ linear position, while the rate gyroscope can determine angular position (i.e., joint angles). Acceleration is the slope of the curve on the velocity versus time graph. Position is calculated via integration from acceleration. The acceleration signal must be integrated twice: once to get velocity and a second time to get position (Roetenberg, Luinge, Baten, & Veltink, 2005). The accelerometer in the unit can also estimate inclination or tilt of the sensor. However, since the accelerometer senses both acceleration and the force of gravity, inclination is only accurate when acceleration is low (Veltink et al., 2001). Sabatini, Martelloni, Scapellato, & Cavallo (2005) tested inertial measurement units over a variety of grades and speeds on a treadmill. Compared to an optical motion capture system, the inertial measurement units provided a relatively accurate method of measuring body segment position and orientation. However, performance seems to degrade at steeper inclines. Values are slightly overestimated while walking at steeper inclines. Additionally, walking speed and incline seem to be slightly underestimated (Sabatini, Martelloni, Scapellato, & Cavallo, 2005).

Additionally, Pfau, Witte, & Wilson (2005) determined that inertial measurement units are accurate to track trunk movement on horses at a variety of speeds.

There are several limitations of the inertial measurement units. One of the most prevalent limitations is drift of the orientation of the sensor which limits the ability to utilize the sensors with high accuracy (Roetenberg, Luinge, Baten, & Veltink, 2005). However, this can be prevented by the use of a magnetometer. The magnetometer is sensitive to the earth's magnetic field and reduces the amount of drift that can occur during utilization of the IMU. The magnetometer recognizes magnetic north and can adjust the unit's offset to ensure a more accurate position/orientation estimate. Correction of drift or offset occurs during a period of zero (or near zero) acceleration. Offset and drift errors that occur during data collection is magnified when the acceleration and angular velocity data is integrated. Noise from skin or surface movement can also become magnified if not filtered (Roetenberg, Luinge, Baten, & Veltink, 2005). Inertial measurement units' accuracy is sensitive to more dynamic movements such as running or jumping. Noise can result from the sensor moving relative to the skin during these types of activities. The accuracy of data output is often dependent on the speed and type of movement as well as the surrounding environment. Ferromagnetic materials significantly affect the accuracy of the IMU to correctly identify position and orientation due to disturbances in the direction and density of the local magnetic field. This, in turn, alters orientation estimations. Use of these sensors on larger objects, such as automobiles, will likely require changes in design or filter as the current design is used mainly on human beings and body segments. Changes in air or body temperature can also cause changes in the temperature of the unit, which can cause drift (Roetenberg et al., 2005). Additionally, experimental setups on variable terrain may not produce highly accurate data due to the fact that there may be a greater amount of noise during this type of walking. Rigid mounting of the sensor is essential to gaining accurate data. This can become a problem in individuals with excess tissue, as there is often more movement of the surface of the skin during walking (Pfau, et al., 2005). Additionally, as incline increases, the accuracy of the IMU seems to decrease slightly (Sabatini et al., 2005). This may be due to increased unit movement at steeper inclines.

Video-Based Motion Capture

Video-based motion capture is the most common method for recording kinematic data and is an alternative to inertial measurement units. Reflective markers are used to identify joint position. These markers are placed on various body landmarks, such as the anterior superior iliac spine, sacrum, greater trochanter, and lateral malleolus. The position of the markers in space is generally tracked automatically via software. These markers are identified via their reflective area, so precise placement of the markers is necessary to obtain accurate data. A minimum of two cameras is needed to document the instantaneous location of each marker and obtain the marker's three dimensional coordinates. However, to truly track all markers throughout the gait cycle, it is necessary to have at least three to five cameras. The additional cameras can account for rotational deviations in the subjects' movement. Cameras must be situated in various parts of the room such that it is possible to view all markers on the subject. However, this type of motion analysis is hindered by events such as a hand or arm swing over a hip marker. Extreme rotation or overlap of marker locations can also cause marker dropout. In the event of this occurrence, it is important to manually locate the marker and its trajectory (Perry, 1992; Winter, 2009; Moeslund & Granum, 2001).

Processing Kinematic Data

IMU's record angular velocity and linear acceleration and this data must be integrated to determine angular or linear position (Griffiths, 2006). Integration (area under the acceleration/velocity vs. time) of data to determine velocity and position requires that noise in the raw data be minimized. Standard video-based motion capture systems record the position of markers and do not require the data to be integrated to obtain position. However, both methods have noise in the raw data. The smaller the signal to noise ratio of the raw data, the larger the error in calculated values (e.g. velocity and position). During the integration process, noise in the data is amplified to such an extent that errors can become a large and obtaining accurate data difficult. In general, human movements occur at relatively low frequencies (less than 7-9 Hz for walking). Low-pass filtering can remove higher frequencies noise after data has been collected. Kinematic data is typically low pass filtered with a 4^{th}-order Butterworth filter with a cutoff frequency of 5-10 Hz. This process filters frequencies above the cutoff frequency. Filtering data is effective as long as the frequencies of the raw data and the noise are different (Griffiths, 2006; Winter, Sidwall, & Hobson, 1974). Setting the cutoff frequency too low will eliminate noise but will also remove some of the signal of interest (Winter, 1990; Griffiths, 2006). Using a higher order filter Butterworth filter results in a sharper cutoff that allows smaller amounts of higher frequency signal in compared to lower order Butterworth filters. Utilizing a zero-lag filter can also reduce phase shift in time. This means the data is filtered in both directions (forward in time and backward in time) (Griffiths, 2006; Winter, Sidwall, & Hobson, 1974).

Another method to reduce noise is to use finite difference smoothing. This can be completed by taking a three-point, five-point, or seven-point moving average. However, the disadvantage of using these calculations is that they do not always provide enough smoothing to allow velocity and acceleration to be calculated. Often, some prior smoothing is required to obtain accurate results (Griffiths, 2006; Wood, 1982). Coordinate data can also be smoothed by fitting a quadratic curve to a smaller subset of the data. This can be used to predict smoothed curve values (Griffiths, 2006; Wood, 1982). Kalman filters can also be utilized when real-time filtering is necessary. These filters have the capability to be adjusted by determining differences between incoming data and model predictions (Wood, 1982).

Comparing Kinematic Methods

Methods to compare the IMU and video data include paired t-tests, regression analysis, intraclass correlation coefficients, and Bland-Altman analyses. Regression analysis involves comparison between two groups of data. To determine whether or not the groups are similar, an 'r' coefficient is given. The more similar the groups are, the closer the coefficient is to 1.0. If there is negative correlation between the two groups, the number is closer to -1.0. Finally, if there is no correlation between the groups, the number is 0.0. If the value is near zero, the value of one number cannot be reasonably anticipated based off of another number. A strong positive or negative correlation can typically be observed via a scatter plot. A perfect linear association is a straight line between all of the points (Kleinbaum, Kupper, Nizam, & Muller, 2008).

Because of the fact that the IMU and video data are two different pieces of equipment used on the same person, we can also utilize intraclass correlation to determine if the two methods are similar or different. The intraclass correlation coefficient can be used to test reliability of an instrument. The values range from 0.0 to 1.0. A value of 0.0 means a low correlation between the two sets of data, while a value of 1.0 means there is a strong correlation between the data sets (Armstrong, 1981).

A final method of analyzing the output between ~~the Xsens~~IMU data and video data is a Bland-Altman plot. This is based on the premise that any two methods of measuring the same parameter should have a high correlation. The x axis shows the mean of the results of both methods being compared and the y axis shows the absolute difference between the two methods. These difference plots are very useful in method-comparison studies. The y axis can be used to show absolute, percent, or logarithmic values depending on the size of the scale (Dewitte, Fierens, Stockl, & Thienpont, 2002). A Bland-Altman plot is useful in determining if the data attained from two different methods of measuring the same variable are comparable (Altman & Bland, 1983).

Lower Extremity Kinematics during Snowshoeing

Lower extremity kinematics have not been previously described for snowshoeing. However, there are several potential similarities that may help to gain insight into the differences between normal overground walking and snowshoe walking. The snowshoes add mass to the foot, which may contribute to changes in kinematics. Packed snow may be similar to walking in sand, which alters kinematics. The snowshoe itself has a rigid tail that extends beyond the heel that likely changes kinematics as well. Step width is also likely to be greater than overground walking as a result of the increased width of the snowshoe compared to the foot.

Alterations in kinematics due to added mass at the foot have been previously explained. The addition of weight (2 kg and 4 kg masses) to the foot results in an increase in stride length and swing time and a decrease in stride rate. Elevation angles also changed with foot loads, resulting in a decrease during stance and an increase during swing, which suggests changes in kinematics (Browning, Modica, Kram, & Goswami, 2007).

Snowshoeing involves walking on a relatively soft and unstable surface. This surface can vary depending on the depth and density of the snow. Walking in packed snow may be similar to walking in soft sand. Lejeune, Willems, & Heglund (1998) measured the differences in gravitational potential energy of the center of mass. Their data suggests that there may be differences in kinematics due to walking in a soft sand surface. Walking at a speed of 1.25 m/s shows that the kinetic energy due to the forward movement of the center of mass is equal to the potential energy of the center of mass. It is also shown that there is a change in potential energy. Since mass and gravity remain constant, there must be a change in height of the center of mass to elicit a change in potential energy. This change in height of the center of mass suggests that there are changes in kinematics.

The snowshoes are attached to the foot via a binding, either a fixed or rotating toe-cord. The rotating toe-cord binding drags the snowshoe on the ground and allows snow to fall off the back of the snowshoe when the foot is lifted during swing. The fixed toe-cord does not allow the snow to fall off the snowshoe as easily, as it springs back to the foot when the foot is lifted off the ground in swing. The rotating toe-cord likely results in changes in kinematics due to the snowshoe dragging on the ground (Dalleck, et al., 2003). It seems probable that heel strike and early stance kinematics may be altered due to the addition of the snowshoe to the foot. The rigid extended tail of the snowshoe likely contributes to changes in kinematics. During heel strike, the heel contacts the deck of the snowshoe, which may force the foot into early plantarflexion. This would result in the foot contacting the ground in a more plantarflexed position than the normal neutral foot in overground locomotion. Additionally, it seems feasible that changes would also occur at the hip and knee during swing due to the extended tail of the snowshoe. The foot likely needs to be lifted higher in an effort to prevent tripping because of the snowshoe attached to the foot. These changes would be exaggerated in deeper snow. Changes in frontal plane kinematics are also likely, as the width of the snowshoes would increase step width. This change in step width may especially occur during slow walking, as stability is decreased during slow walking and increasing step width is a mechanism to increase stability during stance.

Snowshoeing Energetics

A few studies have shown that metabolic cost in increased during snowshoeing (Dalleck, et al., 2003; Connolly, Henkin, & Tyzbir, 1999; Babington, Costill, & Getchell, 1999; Schneider et al., 2001). Potential contributors to this increase in metabolic cost include added mass to the foot, walking in a soft surface, the dragging action of the snowshoe on the ground, and an increased step width. Foot loads of 4 kg increase metabolic rate by 36% from baseline (Browning, Modica, Kram, & Goswami, 2007). It seems plausible that although snowshoes weigh less than 4 kg a piece, metabolic rate would be increased in this activity in part due to the weight of the snowshoes. Browning et al. (2007) state that under normal walking conditions, a walking pattern that minimizes metabolic cost is preferred. However, during foot loading, metabolic rate is higher due to the fact that the neuromuscular system attempts to control foot trajectory at the cost of minimizing energy expenditure.

Furthermore, if snowshoeing results in a more flexed hip and knee due to the addition of the snowshoe to the foot, this would also serve to increase metabolic cost. This deviation (flexed posture) from normal walking would have a significant effect on the metabolic cost of the activity. It is known that walking with a bent hip bent knee gait requires significantly greater oxygen consumption. The average increase in oxygen consumption is approximately 52%. The energy cost of locomotion is greater for bent hip bent knee gait up to about speeds of 2.7 $m \cdot s^{-1}$, where the metabolic cost of bent hip bent knee gait actually becomes less than that of normal gait. The energy required to maintain a bent hip bent knee posture is 0.18 ml/kg/m, compared to 0.05 ml/kg/m for normal upright posture. At all measured speeds between 1 and 2 $m \cdot s^{-1}$, oxygen consumption was significantly greater for bent hip bent knee gait than that of overground walking (Carey & Crompton, 2005). Carey & Crompton (2005) state that the increase in metabolic cost of bent hip bent knee walking may be in part due to an increase in muscle activity when walking with a flexed posture. This is in agreement with previous research, which found that EMG activity was significantly greater in the gluteus maximus, rectus femoris, vastus lateralis, lateral gastrocnemius, and tibialis anterior (Grasso, Zago, & Lacquaniti, 2000). Cross country skiing is another activity that results in a more crouched posture; however, there are many kinematic differences between cross country skiing and snowshoeing. Snowshoeing does not provide the glide phase that characterizes cross country skiing. It is well known, however, that the metabolic cost of cross country skiing is significantly higher than that of overground walking (Ainsworth, et al., 1993). Part of this increase may be due to the flexed posture that is demonstrated in cross country skiing.

Walking on packed snow may be similar to walking on sand. There is a small amount of depression while walking on packed snow and on hard sand. This would particularly be observed at toe off. At toe off, the foot would move backward in an effort to continue forward motion at the end of stance. According to Zamparo, Perini, Orizio, Sacher, & Ferretti (1992), walking on sand increases the metabolic cost 60-200% at speeds above 3 km/hr. Energy is lost during this process as not all potential energy is converted to kinetic energy to be used during walking. Some potential energy is lost to the sand. This energy that is lost increases metabolic rate (Lejeune, Willems, & Heglund, 1998; Zamparo, et al., 1992).

Downhill Walking

The adjustments in kinematics observed during level walking are likely more prominent during downhill walking. This is partly due to the extended rigid tail of the snowshoe. The extended tail of the snowshoe may result in a greater ankle angular velocity and rapid plantarflexion when walking downhill post-heel strike. Redfern & DiPasquale (1997) report greater ankle and knee angular velocities during early stance in downhill walking compared to level walking. Previous research indicates that the knee is significantly more flexed in stance and early swing during downhill walking. Additionally, the hip is more extended during swing during downhill walking (Kuster, Sakurai, & Wood, 1995).

CHAPTER III

METHODS AND PROCEDURES

Subjects

We recruited 12 healthy subjects (6 males, 6 females, 26.5 ± 5.3 years, weight, 67.6 ± 10.7 kg) with previous snowshoe experience to participate in this study. Participants provided written informed consent approved by the Colorado State University Human Research Committee.

Snowshoes

We used two kinds of snowshoes for this study: a standard molded plastic model (conventional, MSR Denali, Cascade Designs, Seattle, WA) and molded plastic model with a flexible tail (flex tail, Tubbs Flex, Tubbs Snowshoe Co, Seattle, WA) (Figure 1). The conventional snowshoes weighed 0.85 kg each and were 0.2 m wide and 0.53 m long and were rigid along their length. We attempted to match the size of the flex tail snowshoe with the conventional, but the size-matched flex tail snowshoe would not accommodate larger foot sizes (> Men's US 9), so we used a larger flex tail snowshoe for the male participants. The flex tail snowshoes weighed 0.78 and 0.85 kg each and were 0.53 m and 0.61 m long, for the small and large snowshoes respectively. Both sizes of the flex tail snowshoes were 0.2 m wide. Both types of snowshoes had longitudinal steel traction blades along the underside of the snowshoe and steel crampons attached to a rotating toe cord binding. This rotating toe cord design (Dalleck, et al., 2003) allows the foot to rotate in the sagittal plane and results in the snowshoe being dragged across the snow during leg swing. The small flex tail snowshoes flexed at 0.08 m from the end of the tail, while the large flex tail snowshoes flexed at 0.10 m from the end of the tail (Figure 1B).

Study Protocol

Each study participant performed eight walking trials, two overground and six on snow. Trial order was randomized for each subject. All trials took place the same day for each subject to minimize differences in snow conditions. Subjects wore tight fitting tights or pants and low rise running or hiking shoes throughout all trials. Two overground level walking trials were conducted without snowshoes at both 3 km $\cdot$ hr^{-1} (0.9 m $\cdot$ s^{-1}) and 5 km $\cdot$ hr^{-1} (1.4 m $\cdot$ s^{-1}). All snowshoe trials were conducted on packed snow that was mechanically machine groomed and provided a small amount (<1 cm) of depression. Four level snowshoe trials were conducted using flex tail and conventional style snowshoes at both 0.9 m $\cdot$ s^{-1} and 1.4 m $\cdot$ s^{-1}. During each level trial, participants walked back and forth five times along a 100 m groomed track. Each participant self-monitored speed via a GPS unit worn around the wrist (Garmin Forerunner 205, Olathe, Kansas). Additionally, one downhill trial were completed by each subject using the two types of snowshoes. Downhill trials were executed at a self-selected speed on a 14° slope of approximately 20 meters in length. During each downhill trial, participants walked downhill and uphill five times.

Kinematics

We recorded the time to complete 10 strides to determine stride frequency and recorded the time required to walk 5 meters to calculate walking speed. We used small (3 x 2.5 cm) lightweight inertial measurement units (IMU, Xsens MTx, Enschede, The Netherlands) to collect lower extremity kinematic data. Each IMU sensor contains a three-dimensional accelerometer, three rate gyroscopes and a magnetometer. These IMU's have the ability to measure tilt and magnetic north to accurately measure three dimensional angular velocity and linear acceleration. Drift of the unit's position and orientation are minimized by the use of a magnetometer that is sensitive to the earth's magnetic field (Roetenberg, et al., 2005). The IMU's were placed on the subject's sacrum, right thigh, shank, and on the dorsal aspect of the right foot (see Figure 2). IMU sensor data was collected at 50 Hz. Euler angle data (roll, pitch and yaw angles) were collected for each sensor and were used to determine joint angles utilizing X-analyzer software (NexGen Ergonomics, Pointe Claire, Quebec). The X-analyzer software used the relative position of the sensors distal and proximal to each joint to calculate three-dimensional hip, knee, and ankle joint angles. Data was collected for 30-60 seconds per trial.

To determine step width and confirm the accuracy of the IMU data, we also recorded frontal and sagittal plane video data during each trial. Video cameras were placed at the end of the 100 m track to record frontal plane movement and at the mid-point of the loop to record sagittal plane movement. Reflective markers were placed on the right side of the body on the anterior superior iliac spine, superior border of the greater trochanter, lateral femoral epicondyle, lateral malleolus, lateral aspect of the fifth metatarsal head, and the posterior portion of the heel. Markers were also placed on the front and rear of the right snowshoe. The sagittal plane video data was digitized and the angles between two adjacent segments (pelvis and thigh, thigh and shank, shank and foot) were computed for a stride using motion capture software (Vicon Motus V9.2, Vicon, Centennial, CO). Knee range of motion (maximum-minimum values) was calculated for each subject and compared to knee range of motion data from the IMUs. Step width was determined by digitizing the medial malleolus of each ankle during mid-stance and calculating the distance between the two markers.

Hip, knee and ankle joint angles for each trial were low-pass filtered at 12 Hz using a 4^{th} order Butterworth filter. We used a custom-designed software program (Matlab version 7.6, Natick, MA) to determine gait cycle events (heel-strike, toe-off) and mean joint angles for each individual trial and the group. We defined heel strike as the point of maximum knee extension after swing (Borghese, et al., 1996). Joint angular velocities were calculated from ankle joint data. We focused specifically on the period from peak knee extension at heel strike to peak knee flexion during stance for overground trials. Joint angular velocities for snowshoe trials were determined from peak velocity during late swing through early stance. We used this part of the gait cycle because ankle plantarflexion began before peak knee extension during the snowshoe trials (Figure 3).

Statistical Analysis

A one-way ANOVA with repeated measures was used to determine how snowshoes and walking speed affected lower extremity kinematics. If significant main effects were observed, post-hoc comparisons using the Holm-Sidak test were performed.

Paired t-tests were used to determine any differences in downhill walking trials. A criterion of $p<0.05$ defined significance. Paired t-tests were used to establish if there were any differences between the Xsens and video data. We also used linear regression to determine the correlation between the IMU and video data.

REFERENCES

Abbott, B.C. & Wilkie, D.R. (1953). The relation between velocity of shortening and the tension-length curve of skeletal muscle. *J. Physiol, 120*, 214-223.

Ainsworth, B.E., Haskell, W.L., Leon, A.S., Jacobs, D.R., Montoye, H.J., Sallis, J.F., & Paffenbarger, R.S. (1993). Compendium of physical activities: classification of energy costs of human physical activities. *Med Sci Sports Exerc*, *25*(1), 71-80.

Altman, D.G., Bland, J.M. (1983). Measurement in medicine: the analysis of method comparison studies. *Statistician, 32*, 307-317.

Alton, F., Baldey, L., Caplan, S., & Morrissey, M. C. (1998). A kinematic comparison of overground and treadmill walking. *Clin Biomech (Bristol, Avon), 13*(6), 434-440.

Armstrong, G.D. (1981). The intraclass correlation as a measure of interrater reliability of subjective judgments. *Nursing Research, 30*(5), 314-315.

Babington, J.P., Costill, D.L., & Getchell, L. (1999). Energy cost of snowshoe exercise. *Med Sci Sports Exerc, 31*(5), S366.

Bigland, B. & Lippold, O.C.J. (1954). The relation between force, velocity, and integrated electrical activity in human muscles. *J. Physiol, 123*, 214-224.

Borghese, N.A., Bianchi, L., & Lacquaniti, F. (1996). Kinematic determinants of human locomotion. *Journal of Physiology 494.3*, 863-869.

Browning, R. C., Modica, J., Kram, R., & Goswami, A. (2007). The Effects of Adding Mass to the Legs on the Energetics and Biomechanics of Walking. *Med Sci Sports Exerc, 39*(3), 515-525.

Carey, T. S., & Crompton, R. H. (2005). The metabolic costs of 'bent-hip, bent-knee' walking in humans. *J Hum Evol, 48*(1), 25-44.

Cavagna, G.A., Heglund, N.C., & Taylor, C.R. (1977). Mechanical work in terrestrial locomotion: two basic mechanisms for minimizing energy expenditure. *Am J Physiol Regulatory Integrative Comp Physiol, 233*, 243-261.

Cavagna, G.A. & Margaria, R. (1966). Mechanics of walking. *J Appl Physiol, 21*(1), 271-278.

Chiu, M. C., & Wang, M. J. (2007). The effect of gait speed and gender on perceived exertion, muscle activity, joint motion of lower extremity, ground reaction force and heart rate during normal walking. *Gait Posture, 25*(3), 385-392.

Connolly, D., Henkin, J.A., & Tyzbir, R.S. The energy expenditure of snowshoeing in packed vs unpacked snow at low level walking speeds. *Med Sci Sports Exerc, 31*(5), S367.

Dalleck, L. C., DeVoe, D. E., & Kravitz, L. (2003). Energy cost and physiological responses of males snowshoeing with rotating and fixed toe-cord designs in powdered snow conditions. *Ergonomics*, *46*(9), 875-881.

Dewitte, K., Fierens, C., Stockl, D., & Thienpont, L.M. (2002). Application of the Bland-Altman plot for the interpretation of method-comparison studies: a critical investigation of its practice. *Clin Chem*, *48*, 799-801.

Donelan, J. M., Kram, R., & Kuo, A. D. (2001). Mechanical and metabolic determinants of the preferred step width in human walking. *Proc Biol Sci*, *268*(1480), 1985-1992.

Fenn, W.O. & Marsh, B.S. (1935). Muscular force at different speeds of shortening. *J. Physiol*, *85*(3), 277-297.

Grasso, R., Zago, M., & Lacquaniti, F. (2000). Interactions between posture and locomotion: motor patterns in humans walking with bent posture versus erect posture. *J Neurophysiol*, *83*(1), 288-300.

Griffiths, I.W. (2006). Principles of biomechanics and motion analysis. Philadelphia: Lippincott Williams & Wilkins.

Hall, C., Figueroa, A., Fernhall, B., & Kanaley, J. A. (2004). Energy expenditure of walking and running: comparison with prediction equations. *Med Sci Sports Exerc*, *36*(12), 2128-2134.

Ivanenko, Y. P., Grasso, R., Macellari, V., & Lacquaniti, F. (2002). Control of foot trajectory in human locomotion: Role of ground contact forces in simulated reduced gravity. *J Neurophysiol, 87*, 3070-3089.

Kirtley, C., Whittle, M.W., & Jefferson, R.J. (1985). Influence of walking speed on gait parameters. *J Biomed Eng, 7*, 282-288.

Kleinbaum, Kupper, Nizam, & Muller. (2008). Applied regression analysis and other multivariable methods. Belmont: Thomson Higher Education.

Kuo, A.D. (2007). The six determinants of gait and the inverted pendulum analogy: a dynamic walking perspective. *Human Movement Science, 26*, 617-656.

Kuster, M., Sakurai, S., & Wood, G. A. (1995). Kinematic and kinetic comparison of downhill and level walking. *Clin Biomech (Bristol, Avon), 10*(2), 79-84.

Lee, C.R. & Farley, C.T. (1998). Determinants of the center of mass trajectory in human walking and running. *The Journal of Experimental Biology, 201*, 2935-2944.

Lejeune, T. M., Willems, P. A., & Heglund, N. C. (1998). Mechanics and energetics of human locomotion on sand. *Journal of Experimental Biology, 201*(13), 2071-2080.

McGeer, T. (1993). Dynamics and control of bipedal locomotion. *J. theor. Biol., 163,* 277-314.

Moeslund, T.B. & Granum, E. (2001). A survey of computer vision-based human motion capture. *Computer Vision and Vision Understanding, 81*(3), 231-268.

Murray, M.P., Drought, A.B., & Kory, R.C. (1964). Walking patterns of normal men. *J Bone Joint Surg Am, 46*, 335-360.

Murray, M.P., Mollinger, L.A., Gardner, G.M., & Sepic, S.B. (1984). Kinematic and EMG patterns during slow, free, and fast walking. *Journal of Orthopaedic Research, 2,* 272-280.

Murray, M. P., Spurr, G. B., Sepic, S. B., Gardner, G. M., & Mollinger, L. A. (1985). Treadmill vs. floor walking: kinematics, electromyogram, and heart rate. *J Appl Physiol, 59*(1), 87-91.

Oatis, C.A. (2004). Kinesiology: the mechanics & pathomechanics of human movement. Philadelphia: Lippincott Williams & Wilkins.

Ogden, C. L., Carroll, M. D., Curtin, L. R., McDowell, M. A., Tabak, C. J., & Flegal, K. M. (2006). Prevalence of overweight and obesity in the United States, 1999-2004. *Jama, 295*(13), 1549-1555.

Outdoor Industry Foundation. (2006). Outdoor recreation participation study. Boulder: Outdoor Industry Foundation.

Perry, J. (1992). Gait analysis: normal and pathological function. Thorofare, NJ: SLACK Incorpated.

Pfau, T., Witte, T. H., & Wilson, A. M. (2005). A method for deriving displacement data during cyclical movement using an inertial sensor. *J Exp Biol, 208*(Pt 13), 2503-2514.

Rassier, D.E., MacIntosh, B.R., & Herzog, W. (1999). Length dependence of active force production in skeletal muscle. *J Appl Physiol, 86,* 1445-1457.

Redfern, M.S. & DiPasquale, J. (1997). Biomechanics of descending ramps. *Gait & Posture, 6*, 119-125.

Roetenberg, D., Luinge, H. J., Baten, C. T., & Veltink, P. H. (2005). Compensation of magnetic disturbances improves inertial and magnetic sensing of human body segment orientation. *IEEE Trans Neural Syst Rehabil Eng, 13*(3), 395-405.

Rose, J. & Gamble, J.G. (2006). Human walking. Philadelphia: Lippincott, Williams, & Wilkins.

Royer, T.D. & Martin, P.E. (2001). Altered motor control strategies during the swing phase of walking with leg loads. *Med Sci Sports Exerc, 33*(5), S152.

Sabatini, A. M., Martelloni, C., Scapellato, S., & Cavallo, F. (2005). Assessment of walking features from foot inertial sensing. *Ieee Transactions on Biomedical Engineering, 52*(3), 486-494.

Schneider, P., Porcari, J., Erickson, J., Foster, C., Brice, G., & Freeman, A. (2001). Physiological responses to recreational snowshoeing. *J Exerc Physiol Online, 4*, 45-52.

Veltink, P.H., Luinge, H.J., Kooi, B.J., Baten, C., Slycke, P., Olthuis, W., & Bergveld, P. (2001). The artificial vestibular system - design of a tri-axial inertial sensor system and its application in the study of human movement. *Xsens.com.*

Wang, W. J., Crompton, R. H., Li, Y., & Gunther, M. M. (2003). Energy transformation during erect and 'bent-hip, bent-knee' walking by humans with implications for the evolution of bipedalism. *J Hum Evol, 44*(5), 563-579.

Westing, S.H., Cresswell, A.G., & Thorstensson, A. (1991). Muscle activation during maximal voluntary eccentric and concentric knee extension. *Eur J Appl Physiol, 62*, 104-108.

Wilkie, D.R. (1950). The relation between force and velocity in human muscle. *J. Physiol., 110*, 249-280.

Winter, D.A. (1979). A new definition of mechanical work done in human movement. *J Appl Physiol, 46*(1), 79-83.

Winter, D.A. (1990). Biomechanics and motor control of human movement. New York: John Wiley & Sons, Inc.

Winter, D.A. (1992). Foot trajectory in human gait: a precise and multifactorial motor control task. *Physical Therapy, 72*(1), 45-53.

Winter, D.A. (1995). Human balance and posture control during standing and walking. *Gait & Posture, 3*, 193-214.

Winter, D.A. (2009). Biomechanics and motor control of human movement. Hoboken: John Wiley & Sons, Inc.

Winter, D.A. & Robertson, D.G.E. (1978). Joint torque and energy patterns in normal gait. *Biol Cybernetics*, *29*, 137-142.

Winter, D.A., Sidwall, H.G., & Hobson, D.A. (1974). Measurement and reduction of noise in kinematics of locomotion. *J. Biomechanics, 7*, 157-159.

Wood, G.A. (1982). Data smoothing and differentiation procedures in biomechanics. *Exercise and Sports Sciences Reviews*, *10*(1), 308-362.

Zamparo, P., Perini, R., Orizio, C., Sacher, M., & Ferretti, G. (1992). The energy cost of walking or running on sand. *European Journal of Applied Physiology*, *65*(2), 183-187.

Original Research

Biomechanics of Walking with Snowshoes

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Abstract

The purpose was to determine the effects of snowshoes on lower extremity kinematics during level and downhill (DH) walking. Hypotheses: 1) lower extremity sagittal plane kinematics would be altered during level snowshoe walking 2) during DH walking, a snowshoe with a flexible tail (FT) frame would reduce plantarflexion and knee flexion angular velocities during early stance. Twelve adults completed 6 level walking trials. Subjects walked on packed snow using conventional snowshoes and FT snowshoes and overground without snowshoes. Subjects completed two DH trials with conventional and FT snowshoes. We used lightweight inertial/gyroscopic sensors to collect kinematic data. Subjects had greater hip and knee flexion during stance and greater hip flexion during swing while snowshoe walking on level ground. Plantarflexion began earlier in the gait cycle during snowshoe walking. Lower extremity kinematics were similar across snowshoe frame designs during level and DH walking.

Introduction

Snowshoeing is a popular winter activity, both as a competitive and recreational sport. The number of individuals who snowshoe continues to increase, reaching ~5 million participants in 2006 (Outdoor Industry Foundation, 2006). Snowshoeing's popularity is due, in part, because it is an easy-to-learn, low-impact recreational activity. In addition, improvements in design have resulted in lightweight snowshoes that are easy to use because they provide adequate traction, flotation, and stability. The metabolic rate during snowshoeing is ~2x greater than during overground walking, making it an excellent form of exercise (Schneider et al., 2001; Hall, Figueroa, Fernhall, & Kanaley, 2004). The increase in metabolic rate during snowshoeing is likely caused by a combination of altered biomechanics, added mass of the snowshoe and walking on a soft surface. However, the biomechanics of walking with snowshoes have not been reported. Quantifying lower extremity biomechanics during snowshoe walking may help explain the greater metabolic rate during snowshoeing. In addition, evaluating how snowshoe frame design affects lower extremity biomechanics may provide guidance for further improvements.

Lower extremity kinematics have not been described during snowshoeing. However, studies that have examined kinematics during level walking combined with body-weight support, added foot mass, or a soft surface, provide some insights into potential kinematic adjustments. The position and trajectory of the foot is tightly controlled during normal (Winter, 1992) and body-weight supported walking (Ivanenko, Grasso, Macellari, & Lacquaniti, 2002), most likely to minimize metabolic cost and to prevent tripping and slipping. The addition of a snowshoe is likely to require adjustments in foot trajectory and therefore lower extremity kinematics. Adding mass to the foot during level, treadmill walking increased stride length and swing time and altered segment elevation angles at toe-off (Browning, Modica, Kram, & Goswami, 2007). Lejeune, Willems, & Heglund (1998) reported the mechanics of walking in loose sand vs. overground. Although Lejeune et al (1998) did not report kinematics, the differences in center of mass gravitational potential energy between walking in sand and overground suggest changes in lower extremity joint angles (Lejeune, Willems, & Heglund, 1998).

It seems reasonable that the addition of a snowshoe to the foot would alter lower extremity kinematics during walking. Most snowshoes are attached to the foot via a binding that allows the snowshoe to rotate around the attachment point, just inferior to the metatarsals. As a result, the snowshoe drags along the snow as the foot swings forward. While this design allows accumulating snow to slide off the back of the snowshoe during leg swing, the extended, rigid tail of the snowshoe may alter lower extremity kinematics, particularly during swing and early stance. As the foot swings forward and the ankle dorsiflexes, the heel typically contacts the deck (top surface) of the snowshoe, which may force the foot into plantarflexion, essentially beginning heel strike earlier in leg swing. Snowshoers may accommodate for the extended heel by lifting the foot higher, using a more flexed hip and knee throughout swing, and keeping the ankle more plantarflexed prior to heel strike. Frontal plane kinematics are also likely to be different during snowshoe vs. overground walking, because of the additional width of the snowshoe frame. The wider snowshoes would be expected to increase step width, especially during slower walking.

The kinematic adjustments caused by walking in snowshoes may become more pronounced when walking downhill, as the extended heel would cause rapid ankle plantarflexion once the heel contacts the snowshoe deck. Downhill walking results in a more flexed posture at the hip, knee and ankle compared to overground level walking. The ankle and knee angular velocities are also greater during early stance in downhill vs. level walking (Redfern & DiPasquale, 1997). A snowshoe design that incorporates a flexible tail may reduce the need to alter kinematics, particularly during downhill walking, as the tail would flex during late swing and early stance.

The purpose of this study was to determine the effects of snowshoes and their frame design on lower extremity kinematics during level and downhill walking. We hypothesized the following: 1) lower extremity sagittal plane kinematics would be different during level snowshoe vs. overground walking and 2) during downhill walking, the use of a snowshoe with a flexible tail frame would reduce ankle plantarflexion and knee flexion angular velocities during early stance.

Methods

Subjects

We recruited 12 healthy subjects (6 males, 6 females, 26.5 ± 5.3 years, weight, 67.6 ± 10.7 kg) with previous snowshoe experience to participate in this study. Participants provided written informed consent approved by the Colorado State University Human Research Committee.

Snowshoes

We used two kinds of snowshoes for this study: a standard molded plastic model (conventional, MSR Denali, Cascade Designs, Seattle, WA) and molded plastic model with a flexible tail (flex tail, Tubbs Flex, Tubbs Snowshoe Co, Seattle, WA) (Figure 1). The conventional snowshoes weighed 0.85 kg each and were 0.2 m wide and 0.53 m long and were rigid along their length. We attempted to match the size of the flex tail snowshoe with the conventional, but the size-matched flex tail snowshoe would not accommodate larger foot sizes (> Men's US 9), so we used a larger flex tail snowshoe for the male participants. The flex tail snowshoes weighed 0.78 and 0.85 kg each and were 0.53 m and 0.61 m long, for the small and large snowshoes respectively. Both sizes of the flex tail snowshoes were 0.2 m wide. Both types of snowshoes had longitudinal steel traction blades along the underside of the snowshoe and steel crampons attached to a rotating toe cord binding. This rotating toe cord design (Dalleck, DeVoe, & Kravitz, 2003) allows the foot to rotate in the sagittal plane and results in the snowshoe being dragged across the snow during leg swing. The small flex tail snowshoes flexed at 0.08 m from the end of the tail, while the large flex tail snowshoes flexed at 0.10 m from the end of the tail (Figure 1B).

Study Protocol

Each study participant performed eight walking trials, two overground and six on snow. Trial order was randomized for each subject. All trials took place the same day for each subject to minimize differences in snow conditions. Subjects wore tight fitting tights or pants and low rise running or hiking shoes throughout all trials. Two overground level walking trials were conducted without snowshoes at both 3 km · hr^{-1} (0.9 m · s^{-1}) and 5 km · hr^{-1} (1.4 m · s^{-1}). All snowshoe trials were conducted on packed snow that was mechanically machine groomed and provided a small amount (<1 cm) of depression. Four level snowshoe trials were conducted using flex tail and conventional style snowshoes at both 0.9 m · s^{-1} and 1.4 m · s^{-1}. During each level trial, participants walked back and forth five times along a 100 m groomed track. Each participant self-monitored speed via a GPS unit worn around the wrist (Garmin Forerunner 205, Olathe, Kansas). Additionally, one downhill trial was completed by each subject using the two types of snowshoes. Downhill trials were executed at a self-selected speed on a 14° slope of approximately 20 meters in length. During each downhill trial, participants walked downhill and uphill five times.

Kinematics

We recorded the time to complete 10 strides to determine stride frequency and recorded the time required to walk 5 meters to calculate walking speed. We used small (3 x 2.5 cm) lightweight inertial measurement units (IMU, Xsens MTx, Enschede, The Netherlands) to collect lower extremity kinematic data. Each IMU sensor contains a three-dimensional accelerometer, three rate gyroscopes and a magnetometer. These IMU's have the ability to measure tilt and magnetic north to accurately measure three dimensional angular velocity and linear acceleration. Drift of the unit's position and orientation are minimized by the use of a magnetometer that is sensitive to the earth's magnetic field (Roetenberg, Luinge, Baten, & Veltink, 2005). The IMU's were placed on the subject's sacrum, right thigh, shank, and on the dorsal aspect of the right foot (see Figure 2). IMU sensor data was collected at 50 Hz. Euler angle data (roll, pitch and yaw angles) were collected for each sensor and were used to determine joint angles utilizing X-analyzer software (NexGen Ergonomics, Pointe Claire, Quebec). The X-analyzer software used the relative position of the sensors distal and proximal to each joint to calculate three-dimensional hip, knee, and ankle joint angles. Data was collected for 30-60 seconds per trial.

To determine step width and confirm the accuracy of the IMU data, we also recorded frontal and sagittal plane video data during each trial. Video cameras were placed at the end of the 100 m track to record frontal plane movement and at the mid-point of the loop to record sagittal plane movement. Reflective markers were placed on the right side of the body on the anterior superior iliac spine, superior border of the greater trochanter, lateral femoral epicondyle, lateral malleolus, lateral aspect of the fifth metatarsal head, and the posterior portion of the heel. Markers were also placed on the front and rear of the right snowshoe. The sagittal plane video data was digitized and the angles between two adjacent segments (pelvis and thigh, thigh and shank, shank and foot) were computed for a stride using motion capture software (Vicon Motus V9.2, Vicon, Centennial, CO). Knee range of motion (maximum-minimum values) was calculated for each subject and compared to knee range of motion data from the IMUs. Step width was determined by digitizing the medial malleolus of each ankle during mid-stance and calculating the distance between the two markers.

Hip, knee and ankle joint angles for each trial were low-pass filtered at 12 Hz using a 4th order Butterworth filter. We used a custom-designed software program (Matlab version 7.6, Natick, MA) to determine gait cycle events (heel-strike, toe-off) and mean joint angles for each individual trial and the group. We defined heel strike as the point of maximum knee extension after swing (Borghese, Bianchi, & Lacquaniti, 1996). Joint angular velocities were calculated from ankle joint data. We focused specifically on the period from peak knee extension at heel strike to peak knee flexion during stance for overground trials. Joint angular velocities for snowshoe trials were determined from peak velocity during late swing through early stance. We used this part of the gait cycle because ankle plantarflexion began before peak knee extension during the snowshoe trials (Figure 3).

Statistical Analysis

A one-way ANOVA with repeated measures was used to determine how snowshoes and walking speed affected lower extremity kinematics. If significant main effects were observed, post-hoc comparisons using the Holm-Sidak test were performed. Paired t-tests were used to determine any differences in downhill walking trials. A criterion of $p<0.05$ defined significance. Paired t-tests and linear regression were used to establish if there were any differences between the Xsens and video data.

Results

In general, temporal gait characteristics were similar during overground walking vs. snowshoeing at each speed (Table 1). During the slow walking trials, participants walked slightly faster than the target speed of $0.9 \text{ m} \cdot \text{s}^{-1}$, but there were no significant differences in speed between the trials ($p=0.679$). At this slow speed, there were no differences in stride length ($p=0.467$) and stride frequency ($p=0.684$) across the overground and on-snow trials. When asked to maintain a speed of $1.4 \text{ m} \cdot \text{s}^{-1}$, participants walked slightly faster overground and slower on snow. However, the differences in speed were not significant (p=0.112). Due to the slower on-snow speed, stride frequency was reduced during snowshoe walking compared to overground walking (p=0.009). As expected, both stride length and stride frequency increased as walking speed increased on level ground during both snowshoe and overground walking. Participants walked downhill at ~1.1 m · s^{-1}. Step width was greater during snowshoe than overground walking at 0.9 m · s^{-1} (p= 0.012,). Step width at 1.4 m · s^{-1} was not significantly different between the snowshoe walking and overground trials (p=0.065).

There were several differences in lower extremity joint angles when comparing level overground vs. and snowshoe walking, particularly at the faster walking speed. Hip joint angles were similar across the trials when walking at 0.9 m · s^{-1}, but at the faster walking speed the hip was more flexed during early stance and swing while snowshoeing when compared to overground walking (Figure 3 A, D). Participants also walked with a more flexed knee angle during early stance in snowshoe vs. overground walking (p=<0.001) (Figure 3 B, E). Although the knee was more flexed during early stance in level snowshoeing vs. overground walking, knee range of motion and time-to-peak flexion during the first half of stance were similar across the trials. As a result, mean and peak knee flexion angular velocities during early stance were similar across the trials (Figure 4). Ankle plantarflexion began during swing while snowshoeing but began at heel strike while walking overground (Figure 3 C, F). In addition, peak ankle plantarflexion angular velocity was significantly greater during overground vs. snowshoe walking (p=0.019) (Figure 4). There were no differences in lower extremity kinematics in the conventional and flex-tail snowshoe designs.

Although there were differences in joint angles between downhill and level snowshoeing, no significant differences between the conventional and the flex tail snowshoes during downhill walking were detected. At heel strike, the hip was slightly more extended during downhill than level walking (Figure 5A). Knee angle during midstance and early swing was more flexed when snowshoeing downhill. During late stance and early swing the ankle had a greater amount of dorsiflexion. No significant differences existed between the two snowshoe designs for mean and peak joint angular velocities.

The inertial measurement unit output was compared to a more established technique using video data. The difference between the two was not statistically significant, suggesting that the IMU's are a valid method of assessing kinematic data. The r^2 value for our linear regression analysis was 0.557, thus there was a correlation between the two methods.

Discussion

The results of this study show that individuals modified their gait by altering lower-extremity joint angles when walking on level, packed snow using snowshoes compared to overground walking, particularly at faster walking speeds. Thus, we accept our hypothesis that kinematics are different for snowshoeing vs. overground walking. When participants walked downhill on snowshoes, the hip and knee were more flexed during mid-late stance and the ankle was more dorsiflexed during late stance compared to snowshoeing on level ground. No significant differences were observed between the snowshoe frame designs. Additionally, step width tended to be wider when walking with snowshoes vs. overground.

Hip and knee joint angle patterns were similar during snowshoeing vs. overground walking. However, the ankle joint angle pattern during late stance and swing was different across the two forms of locomotion. Our overground kinematic data is similar to those reported by others (Chiu & Wang, 2007; Rose & Gamble, 2006). When walking at 1.4 $m \cdot s^{-1}$ in snowshoes, subjects walked with a more flexed hip and knee during the first half of stance and throughout swing. Ankle joint kinematics were distinctly different during snowshoeing vs. overground walking. While overall ranges of motion were similar, the pattern was shifted such that the ankle plantarflexion associated with heel strike during overground walking occurred during swing when snowshoeing. This is similar to a “shuffle” style gait where the foot is held in a more horizontal position during swing. This is likely due to the tail of the snowshoe protruding posterior to the heel, which would prevent dorsiflexion prior to heel strike and may result in a greater degree of flexion at the knee during early stance. The crouched or “bent-hip bent-knee” gait may be an accommodation to walking on an unstable surface as the flexed posture lowers the center of mass and may improve stability (Wang, Crompton, Li, & Gunther, 2003).

Generally, joint angular velocities were similar between overground walking and snowshoe walking. However, peak ankle plantarflexion angular velocity in late swing and early stance was slower walking on level ground in snowshoes compared to overground walking. We found that the peak ankle plantarflexion angular velocity occurred during late swing while snowshoeing but during early stance while walking overground. The gradual ankle plantarflexion during swing results in a relatively horizontal foot position prior to heel strike; eliminating the need for rapid plantarflexion after heel strike. Conversely, maximum ankle angular velocity occurred during early stance in overground walking, due to a pronounced heel strike (and more dorsiflexed foot) and subsequent rapid plantarflexion.

Measuring joint angular velocities is important because it may be related to the effort and/or comfort an individual experiences during snowshoeing. The action of plantarflexion during early stance is largely controlled by eccentric contraction of the tibialis anterior. The ground reaction force acts posterior to the ankle and creates a plantarflexor torque about this joint. In order to avoid the foot slapping the ground, the tibialis anterior has to eccentrically contract and controls the rate of plantarflexion. A greater angular velocity at heel strike would increase the eccentric shortening velocity of the tibialis anterior, which would increase the force output (assuming the level of activation is similar). Additionally, increased angular velocity may lead to increased power absorption at the ankle. If the ground reaction force, which, during overground walking, is normally under the heel at heel strike, were located behind the heel under the tail of the snowshoe, this would increase the plantarflexor torque at the ankle (greater moment arm). A greater torque and angular velocity would increase power absorption and also the negative work (integral of power vs. time) done by the ankle dorsiflexor muscles. The increase in muscle force, power absorption and negative work may be perceived as less comfortable than level walking and may also increase the risk of musculoskeletal injury compared to level walking.

Our kinematic data suggests that there are differences between level and downhill snowshoeing. The hip and knee were more flexed during mid-late stance and the ankle was more dorsiflexed during late stance compared to snowshoeing on the level. Comparing our data with that of others who have reported kinematics during downhill walking suggests that the knee is more flexed while snowshoeing (Kuster, Sakurai, & Wood, 1995). The increased knee flexion observed during downhill snowshoe walking is likely due to the attachment of snowshoes to the feet and the need to clear the snowshoe from the ground. The pattern of gradual ankle plantarflexion during swing during level snowshoeing is also observed while going downhill. Thus, individuals tended to “slide” the foot forward and eliminate the need for rapid plantarflexion at heel contact. This explains the modest reduction in plantarflexion angular velocity when using the flexible tail snowshoe.

The kinematic differences between snowshoe walking and overground walking provide insights into why the metabolic rate is ~2x greater during snowshoe walking vs. overground walking (Hall, et al., 2004; Schneider, et al., 2001). The increased hip and knee flexion, soft surface, increased step width and walking with weight attached to the feet all likely contribute to the greater metabolic rate observed during snowshoeing. Our data shows that snowshoe walking is characterized by more flexion at the hip and knee. Grasso et al. (2000) showed that this “crouch” style gait observed during snowshoeing results in an increase in metabolic rate due to a higher level of muscle activity (Grasso, Zago, & Lacquaniti, 2000). The metabolic rate associated with walking at $1.5\ m \cdot s^{-1}$ during bent hip bent knee walking is approximately 45% greater than normal walking (Carey & Crompton, 2005). Although our subjects walked on packed snow, we still observed a small amount of snow depression, specifically at toe off. Thus, snowshoe walking may be similar to walking in hard sand where the foot moves sand backwards in an effort to push the body forward at the end of stance. Thus, energy is lost and metabolic rate is increased when walking on this type of surface (Lejeune, et al., 1998; Zamparo, Perini, Orizio, Sacher, & Ferretti, 1992a). Step width was slightly greater during snowshoe walking and an increase in step width is known to increase metabolic rate (Donelan, Kram, & Kuo, 2001). Finally, metabolic rate during snowshoeing may be increased due to the additional weight attached to the feet. Each snowshoe weighs between 0.80 and 0.85 kg, which would increase in energy expenditure by ~15% based on leg loading studies (Browning, et al., 2007).

In addition to assessing the differences between walking with snowshoes and overground walking, we compared two different snowshoe designs. We found no significant differences in lower extremity kinematics between the two snowshoe designs. The similarity in the results was due, in part, to the length of the snowshoes used in this study. We found that the conventional and small flex tail snowshoes had a shorter tail length (end of the heel to end of the snowshoe) and that tail length was positively related to maximum plantarflexion angular velocity (Figure 6). Thus, tail length is one reason why we did not observe differences between the two designs. The large flex tail snowshoes have a longer heel to tail distance, which increased maximum plantarflexion angular velocity. It may be that differences between the two designs would appear while walking down steeper grades, which are common during snowshoeing. A number of improvements have been made to snowshoes in recent decades. These developments increase the flotation, traction, and stability of the snowshoe. Our data supports the development of snowshoes designed for packed snow conditions that are lightweight, narrow and have a shorter, more flexible tail so that snowshoers can walk more normally on level and downhill terrain. A longer tail results in a higher rate of plantarflexion, especially during downhill walking. Since the flexible tail snowshoe utilized in this study did not have an extreme amount of flexion, it seemingly acted very similar to the conventional snowshoe.

The inertial measurement units (IMUs) used in this study are well-suited for measuring kinematics in a challenging outdoor environment as they are lightweight, easy to use and can be shielded from adverse weather. We found that the kinematics recorded with the IMU's were similar to those measured using standard video motion capture. However, these units do have limitations. The sensors were affixed to clothing, thus, there is a potential for movement artifact. This was minimized by tightly wrapping the sensors to the body segment via elastic athletic tape and filtering the joint angle data. As Sabatini et al.~~,~~ (2005) noted, IMU movement is probably most likely to occur at ground impact. Thus, while running may introduce movement artifact errors, walking probably does not.

Walking with snowshoes on packed snow alters lower extremity kinematics compared to overground walking at similar speeds. The snowshoe gait is characterized by a more flexed posture during stance and a greater degree of plantarflexion during swing. This "shuffling" gait likely contributes to the increased metabolic rate during snowshoeing and was also observed during downhill walking. A snowshoe frame with a flexible vs. rigid tail did not affect kinematics during downhill walking but may reduce the peak plantarflexion angular velocity during level walking and thus make snowshoeing more like overground walking.

Acknowledgements

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References

Borghese, N. A., Bianchi, L., & Lacquaniti, F. (1996). Kinematic determinants of human locomotion. *Journal of Physiology, 494*(3), 863-879.

Browning, R. C., Modica, J., Kram, R., & Goswami, A. (2007). The Effects of Adding Mass to the Legs on the Energetics and Biomechanics of Walking. *Med Sci Sports Exerc, 39*(3), 515-525.

Carey, T. S., & Crompton, R. H. (2005). The metabolic costs of 'bent-hip, bent-knee' walking in humans. *J Hum Evol, 48*(1), 25-44.

Chiu, M. C., & Wang, M. J. (2007). The effect of gait speed and gender on perceived exertion, muscle activity, joint motion of lower extremity, ground reaction force and heart rate during normal walking. *Gait Posture, 25*(3), 385-392.

Dalleck, L. C., DeVoe, D. E., & Kravitz, L. (2003). Energy cost and physiological responses of males snowshoeing with rotating and fixed toe-cord designs in powdered snow conditions. *Ergonomics*, *46*(9), 875-881.

Donelan, J. M., Kram, R., & Kuo, A. D. (2001). Mechanical and metabolic determinants of the preferred step width in human walking. *Proc Biol Sci, 268*(1480), 1985-1992.

Grasso, R., Zago, M., & Lacquaniti, F. (2000). Interactions between posture and locomotion: motor patterns in humans walking with bent posture versus erect posture. *J Neurophysiol, 83*(1), 288-300.

Hall, C., Figueroa, A., Fernhall, B., & Kanaley, J. A. (2004). Energy expenditure of walking and running: comparison with prediction equations. *Med Sci Sports Exerc, 36*(12), 2128-2134.

Ivanenko, Y. P., Grasso, R., Macellari, V., & Lacquaniti, F. (2002). Control of foot trajectory in human locomotion: Role of ground contact forces in simulated reduced gravity. *J Neurophysiol, 87*, 3070-3089.

Kuster, M., Sakurai, S., & Wood, G. A. (1995). Kinematic and kinetic comparison of downhill and level walking. *Clin Biomech (Bristol, Avon), 10*(2), 79-84.

Lejeune, T. M., Willems, P. A., & Heglund, N. C. (1998). Mechanics and energetics of human locomotion on sand. *Journal of Experimental Biology, 201*(13), 2071-2080.

Luinge, H.J. (2002). Inertial sensing of human movement.

Pfau, T., Witte, T. H., & Wilson, A. M. (2005). A method for deriving displacement data during cyclical movement using an inertial sensor. *J Exp Biol, 208*(Pt 13), 2503-2514.

Redfern, M.S. & DiPasquale, J. (1997). Biomechanics of descending ramps. *Gait and Posture, 6*, 119-125.

Roetenberg, D., Luinge, H. J., Baten, C. T., & Veltink, P. H. (2005). Compensation of magnetic disturbances improves inertial and magnetic sensing of human body segment orientation. *IEEE Trans Neural Syst Rehabil Eng, 13*(3), 395-405.

Rose, J. & Gamble, J.G. (1994). Human Walking. Lippincott Williams & Wilkins

Sabatini, A. M., Martelloni, C., Scapellato, S., & Cavallo, F. (2005). Assessment of walking features from foot inertial sensing. *Ieee Transactions on Biomedical Engineering, 52*(3), 486-494.

Schneider, P., Porcari, J., Erickson, J., Foster, C., Brice, G., & Freeman, A. (2001). Physiological responses to recreational snowshoeing. *J Exerc Physiol Online, 4*, 45-52.

Wang, W. J., Crompton, R. H., Li, Y., & Gunther, M. M. (2003). Energy transformation during erect and 'bent-hip, bent-knee' walking by humans with implications for the evolution of bipedalism. *J Hum Evol, 44*(5), 563-579.

Zamparo, P., Perini, R., Orizio, C., Sacher, M., & Ferretti, G. (1992). The energy cost of walking or running on sand. *European Journal of Applied Physiology, 65*(2), 183-187.

Figure Legends

Figure 1: Snowshoes used in this study (A). Shown left to right are the small flex tail, large flex tail, and conventional snowshoes. The small flex tail snowshoe and conventional snowshoe were 0.2 m wide and 0.53 m in length and the large flex tail snowshoe was 0.2 m wide and 0.61 m in length. The flex tail snowshoe had a flexible tail section (B) compared to the rigid conventional snowshoe.

Figure 2: IMU sensor placement on posterior thigh and posterior shank (A). Inset: Xsens MTx IMU that was used to record segment orientation.

Figure 3. Mean joint angles vs. percent stride for the hip, knee, and ankle joints during level overground and snowshoe walking at $0.9\ m \cdot s^{-1}$ (A-C) and $1.4\ m \cdot s^{-1}$ (D-F). Walking with snowshoes resulted in more flexed hip and knee angles, as well as a more dorsiflexed foot.

Figure 4. Mean and peak knee and peak ankle angular velocity during early stance during level overground and snowshoe walking at $1.4\ m \cdot s^{-1}$. Peak ankle angular velocity during snowshoe walking was significantly less than that of overground walking. There we no significant differences between the snowshoe frame designs in mean and peak angular velocities.

Figure 5. Mean joint angles vs. percent stride during downhill walking. C = conventional, DH = downhill, FT = flex tail. The hip (A) and knee (B) had a greater degree of flexion during stance in downhill vs. level snowshoe walking. The ankle (C) had a larger amount of flexion during late stance and early swing during downhill vs. level snowshoe walking.

Figure 6. Relationship between maximum ankle angular velocity and distance between the heel of the shoe and the end of the tail of the snowshoe. C = Conventional, DH = downhill, FT = flex tail. In conventional snowshoes (triangle), an increased heel to tail distance resulted in greater peak ankle angular velocity during downhill walking (long dashed line) ($y=2.736.9x - 153.2$, $r^2=0.41$). Peak angular velocity did not tend to increase with increasing heel to tail distance in flex tail snowshoes (solid line) ($y= -588.9x + 303.6$, $r^2=0.15$).

Table 1. Mean (SD) across all subjects for speed, stride frequency, stride length, and step width.

	Conv 0.9 m/s	**FT 0.9 m/s**	**OG 0.9 m/s**	**Conv 1.4 m/s**	**FT 1.4 m/s**	**OG 1.4 m/s**
Speed (m/s)	0.91 (0.08)	0.93 (0.08)	0.94 (0.08)	1.35 (0.12)	1.35 (0.04)	1.43 (0.11)
Stride Frequency (Hz)	0.76 (0.09)	0.75 (0.07)	0.77 (0.05)	0.92 (0.05)*	0.92 (0.04)*	0.95 (0.05)
Stride Length (m)	1.20 (0.10)	1.25 (0.08)	1.23 (0.08)	1.48 (0.15)	1.48 (0.07)	1.51 (0.12)
Step Width (m)	0.14 (0.04)*	0.16 (0.03)*	0.08 (0.05)	0.13 (0.03)	0.10 (0.03)	0.08 (0.04)

Conv = Conventional, FT = Flex Tail, OG = Overground
*=significantly different from overground ($p<0.05$)

Figure 1.

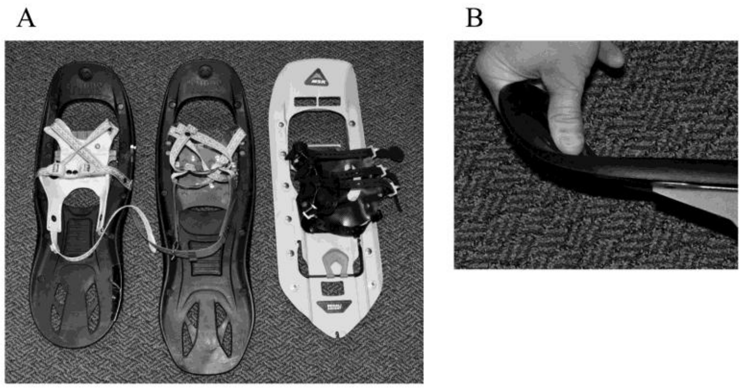
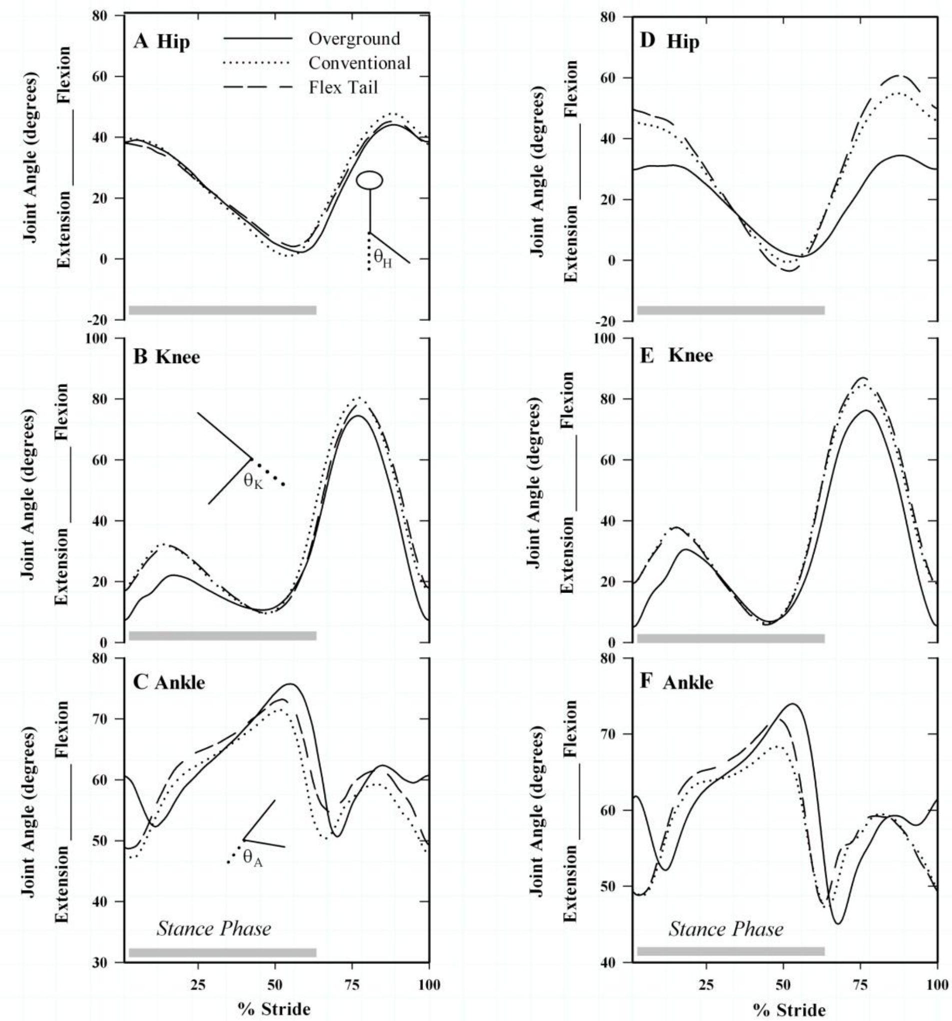


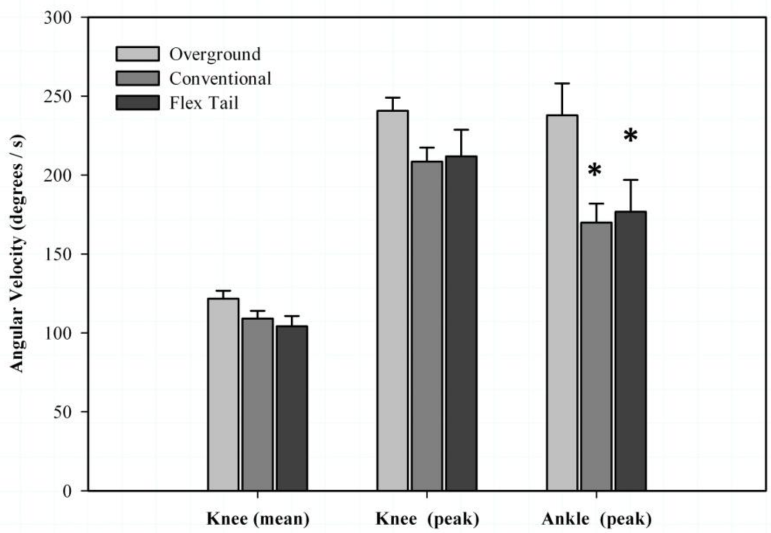
Figure 2.

Figure 3.



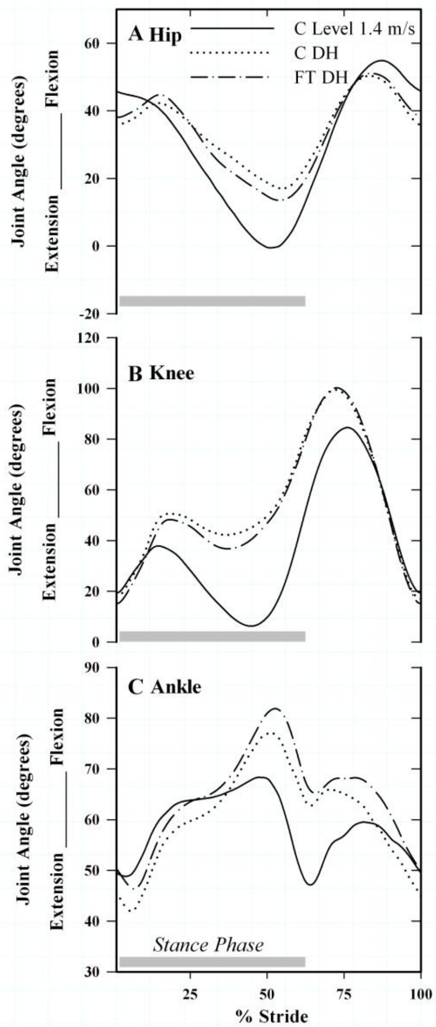
A Hip
Overground
Conventional
Flex Tail
Joint Angle (degrees)
Extension
Flexion
θ_H
B Knee
θ_K
C Ankle
θ_A
Stance Phase
% Stride
D Hip
E Knee
F Ankle

Figure 4.



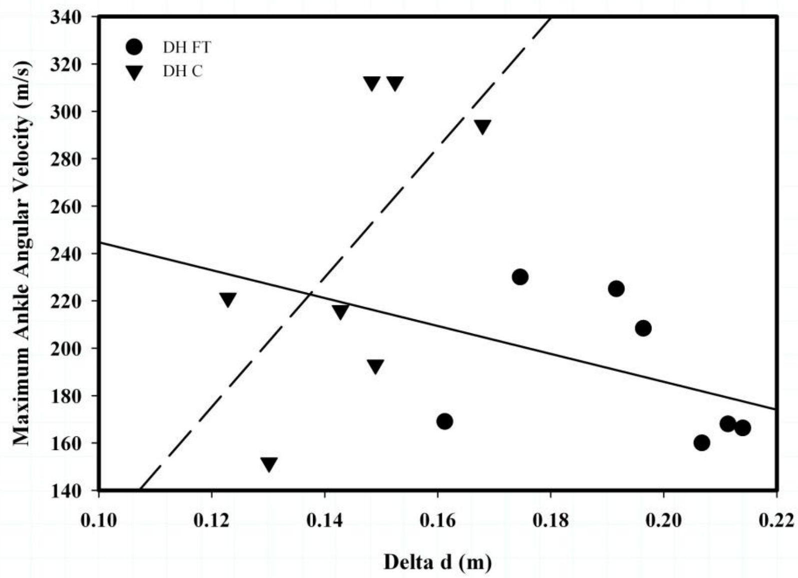
300
250
200
150
100
50
0
Overground
Conventional
Flex Tail
Angular Velocity (degrees / s)
*
*
Knee (mean)
Knee (peak)
Ankle (peak)

Figure 5.



A Hip
C Level 1.4 m/s
C DH
FT DH
Joint Angle (degrees)
Extension
Flexion
60
40
20
0
-20
B Knee
120
100
80
60
40
20
0
C Ankle
90
80
70
60
50
40
30
Stance Phase
25
50
75
100
% Stride

Figure 6.



DH FT
DH C
Maximum Ankle Angular Velocity (m/s)
340
320
300
280
260
240
220
200
180
160
140
0.10
0.12
0.14
0.16
0.18
0.20
0.22
Delta d (m)

APPENDIX A

INFORMED CONSENT FORM

Consent to Participate in a Research Study
Colorado State University

TITLE OF STUDY: The Effects of Snowshoe Frame Design on the Biomechanics and Energetics of Walking with Snowshoes

PRINCIPAL INVESTIGATOR: Ray Browning, PhD. 970-491-5868

WHY AM I BEING INVITED TO TAKE PART IN THIS RESEARCH? You are a physically active man or woman between the ages of 18-45 years and you do not have any major health problems and are not pregnant. Our research is whether the design of a snowshoe affects how you walk and how much energy is required when walking with snowshoes on level, packed snow.

WHO IS DOING THE STUDY? This research is being performed by Ray Browning, Ph.D., of the Health and Exercise Science Department. Trained graduate students, undergraduate students, research associates, or research assistants are assisting with the research. These studies are paid for by K2 Corporation.

WHAT IS THE PURPOSE OF THIS STUDY? Snowshoes are designed to provide flotation, stability and traction to make it easier to walk on snow. The shape and stiffness of the snowshoe may affect how well it provides these benefits. The purpose of this study is to determine if the snowshoe design affects how you walk and how much energy you use when walking on snow.

WHERE IS THE STUDY GOING TO TAKE PLACE AND HOW LONG WILL IT LAST? This study will take place on packed trails at a cross-country ski center and will take approximately 2 hours of your time (not including getting to/from the testing location).

WHAT WILL I BE ASKED TO DO? During the study you will be asked to walk on packed snow using snowshoes and overground without snowshoes. To be sure that you are familiar with walking on snowshoes, you will walk for 15-20 minutes at a comfortable pace before we begin data collection. During the data collection period, you will perform six (6), six-minute trails on level snow/ground and three (3), two-minute trials on gradual downhill snow/ground. You will be allowed to rest between trials. Six of the trials will be on snow using snowshoes and three of the trials will be overground. You will walk at either a moderate or moderately fast speed during each trail. To measure your movements and how much energy you are using, you will have small sensors attached to each foot, lower leg and thigh as well as your low back and you will be wearing a mask that will sample the air that you breathe. The entire experiment should take about 2 hours.

ARE THERE REASONS WHY I SHOULD NOT TAKE PART IN THIS STUDY?
If you are not 18-45 years of age, are pregnant, are a regular smoker, have a condition that limits your ability to walk or run (e.g. knee pain during walking), or have any diseases that would affect our measurements, we will not be able to include you in the research.

 Participant’s initials__________date__________

WHAT ARE THE POSSIBLE RISKS AND DISCOMFORTS?

- Snowshoe/Overground Walking: there is a slight risk you could fall and injure yourself. In addition, you may experience fatigue, shortness of breath, or muscle soreness during or after the walking trials. If you sustain an injury that requires immediate medical attention, we will make sure you receive it. You will be responsible for paying any medical expenses you incur as a result of participating in this study. If you have health insurance, it may pay your medical bills related to an injury sustained during this study and we recommend contacting your health insurance provider to determine if this is the case prior to participating.
- Instrumentation: the devices used to measure kinematics and metabolic rate are non-invasive and pose no known risk.
- It is not possible to identify all potential risks in research procedures, but the researcher(s) have taken reasonable safeguards to minimize any known and potential, but unknown, risks.

ARE THERE ANY BENEFITS FROM TAKING PART IN THIS STUDY? There are no direct benefits to you for participating in this study except the knowing how many calories you burn with walking with and without snowshoes.

DO I HAVE TO TAKE PART IN THE STUDY? Your participation in this research is voluntary. If you decide to participate in the study, you may withdraw your consent and stop participating at any time without penalty or loss of benefits to which you are otherwise entitled.

WHAT WILL IT COST ME TO PARTICIPATE? There is no cost to you for participating except that associated with your transportation to our testing location.

WHO WILL SEE THE INFORMATION THAT I GIVE?
We will keep private all research records that identify you, to the extent allowed by law.

Your information will be combined with information from other people taking part in the study. When we write about the study to share it with other researchers, we will write about the combined information we have gathered. You will not be identified in these written materials. We may publish the results of this study; however, we will keep you name and other identifying information private.

We will make every effort to prevent anyone who is not on the research team from knowing that you gave us information, or what that information is. For example, your name will be kept separate from your research records and these two things will be stored in different places under lock and key. You should know, however, that there are some circumstances in which we may have to show your information to other people. For example, the law may require us to show your information to a court.

CAN MY TAKING PART IN THE STUDY END EARLY? Your participation in the study could end in the rare event of an injury or if you become pregnant.

WILL I RECEIVE ANY COMPENSATION FOR TAKING PART IN THIS STUDY? You will receive a complimentary pair of snowshoes upon completion of the study.

WHAT HAPPENS IF I AM INJURED BECAUSE OF THE RESEARCH? The Colorado Governmental Immunity Act determines and may limit Colorado State University's legal responsibility if an injury happens because of this study. Claims against the University must be filed within 180 days of the injury.

WHAT IF I HAVE QUESTIONS?

Before you decide whether to accept this invitation to take part in the study, please ask any questions that might come to mind now. Later, if you have questions about the study, you can contact the investigator, Ray Browning at 970-491-5868. If you have any questions about your rights as a volunteer in this research, contact Janell Barker, Human Research Administrator at 970-491-1655. We will give you a copy of this consent form to take with you.

Your signature acknowledges that you have read the information stated and willingly sign this consent form. Your signature also acknowledges that you have received, on the date signed, a copy of this document containing 3 pages.

____ _______________________

Signature of person agreeing to take part in the study Date

____

Printed name of person agreeing to take part in the study

____ _______________________

Name of person providing information to participant Date

____

Signature of Research Staff

Page 3 of 3 Participant's initials________date_________