Effects of Frequency on Single Leg Hopping in Typically Developing Preadolescents

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The Effects of Frequency on Single Leg Hopping in Typically Developing Preadolescents

A THESIS

Submitted on Monday July 28, 2014 to the Department of Kinesiology and Health in the College of Education, Georgia State University

In partial fulfillment of the requirements for the degree of Master of Science in Exercise Science, with a Biomechanics Concentration

By

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Abstract

Hopping is considered a mass-spring model movement in which the leg supports the center of mass. There is a preferred hopping frequency and hopping outside of that frequency is more difficult and requires more energy. Leg stiffness has been shown to be an important factor when hopping at different frequencies in young adult populations. The purpose of this study was to observe how a still-developing preadolescent population would modify leg stiffness while hopping at different frequencies and if they have similar motor control strategies compared to young adults. The subjects hopped on their dominant leg to the beat of a metronome at one of four frequency conditions based on their calculated preferred frequency, MP (preferred frequency), MM (20% increase), MF (40% increase), and MS (20% decrease). It was found that this population could change their hopping frequency and they achieved this by manipulating their leg stiffness. At the higher frequency conditions there was less movement of the toe and the center of mass in both the vertical and horizontal directions, including decreased hopping height, decreased COM displacement and COM range of motion. Preadolescents demonstrated an adult-like ability to increase leg stiffness and modulate movement of the toe and the COM while adapting to a range of hopping frequencies. This ability could translate into other mass-spring model movements such as running and jumping.

Keywords: children, hopping, frequency, leg stiffness, motor strategy, center of mass

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Introduction

Hopping, described as unperturbed vertical rebounding off of a singular foot, is a simple movement that follows the constraints of a mass-spring model (C. T. Farley, Blickhan, Saito, & Taylor, 1991). Hopping at a preferred frequency requires little energy because of the mass-spring relationship of the leg (C. T. Farley et al., 1991). In order to maintain this efficient frequency, or change tempo, the nervous system must use a multitude of motor control strategies to effectively stiffen the hopping leg. The specifics of these motor strategies have been well-documented in the healthy adult population (Auyang, Yen, & Chang, 2009; Chang, Roiz, & Auyang, 2008; Claire T. Farley, Houdijk, Strien, & Louie, 1998; Claire T. Farley & Morgenroth, 1999; Yen & Chang, 2010). These previous research efforts would provide the basis of comparison with how a typically developing pre-adolescent age group might coordinate similar or different strategies. Evaluating the pre-adolescent motor strategies would provide useful insight into the development of these motor strategies. This information will also provide a beneficial baseline when studying the development of the same motor skills in special populations.

The Statement of the Question

This project will evaluate the motor control strategies of single leg hopping performance in typically developing pre-adolescents. We will focus on the ability to match a designated frequency, maintain effective leg stiffness, limit toe displacement, and the control of the center of mass.

The Rationale
Hopping is the simplest form of movements that follow a mass-spring model (Hobara, Kanosue, & Suzuki, 2007). Other movements that follow a mass-spring model include running and jumping (Hobara et al., 2007). A mass-spring model involves a spring, in the case of hopping the leg, and a mass, the center of mass (C. T. Farley et al., 1991). Based on the model, when the mass compresses the spring energy is stored and can be used to propel the mass. The most efficient use of this stored energy is found at the preferred frequency during hopping (C. T. Farley et al., 1991). The preferred frequency of the young adult population has been found to be within the range of 2.0-2.2 hops/second (Austin, Tiberio, & Garrett, 2002; C. T. Farley et al., 1991; Ferris & Farley, 1997; Jones & Watt, 1971). Hopping outside of the preferred frequency increases the difficulty of the movement, puts greater demand on the musculoskeletal system, and costs more energy (Auyang et al., 2009). For these reasons, changing the frequency of hopping requires the subject to change the mechanics of the movement in order to maintain performance success (Austin, Tiberio, & Garrett, 2003). Manipulating leg stiffness has been shown to be important in hopping at different frequencies in a young adult population (Austin et al., 2002, 2003; C. T. Farley et al., 1991; Moran, Foley, Parker, & Weiss, 2013).

Leg stiffness in the mass-spring model refers to the proportionality constant of the spring in the system (Butler, Crowell III, & Davis, 2003). The bones, muscles, muscle force, nervous system control, ligaments, cartilage and tendon all contribute to the proportionality constant or stiffness of the leg (Butler et al., 2003). The ability to control the stiffness of the leg is important for movement performance, as well as injury prevention. Increased leg stiffness while hopping contributes to increased hopping frequency and hopping height (Arampatzis, Schade, Walsh, & Brüggemann, 2001; Austin et al., 2002, 2003; C. T. Farley et al., 1991; Granata, Wilson, & Padua, 2002; Moran et al., 2013). While running, increased leg stiffness results in increased
velocity, decreased stride length and decreased energy requirement (Butler et al., 2003; Claire T. Farley & González, 1996). Too much leg stiffness can increase the risk for bony injuries while too little leg stiffness can increase the risk for soft tissue injuries (Butler et al., 2003). It is important for an individual to change their leg stiffness effectively to be able to vary their movement and limit their risk of injury.

The pre-adolescent population is still developing motor control strategies that will be used as adults. Specific to hopping, after the age of six this population has sufficient coordination and development to complete the task and reproduce consistent results (Lang, Busche, Rakhimi, Rawer, & Martin, 2013). A study by Lang et al. (2013) further showed that the ability to maximally stiffen the leg increases with age after seven years old. They found a linear increase in whole body stiffness with age, when normalized to body mass. These findings demonstrate how the preadolescent population is still developing and potentially adapting motor control strategies. Gender did not play a role in their results until age 15 (Lang et al., 2013), which is an age that is older than pre-adolescence and will not be included in this study. Studying how preadolescents adapt to hopping at different frequencies will demonstrate the potential presence and use of leg stiffness control that can be applied to other mass-spring model movements.

**The Hypotheses**

1. Preadolescents will be able to match the metronome at frequencies outside of their preferred but will have greater difficulty with the slowest and fastest frequencies.

2. As frequency increases, the effective leg stiffness will increase and as frequency decreases, the effective leg stiffness will decrease.
3. As frequency increases, the movement of the toe marker in the anterior/posterior and medial/lateral directions will increase.

4. As frequency increases, the movement of the center of mass in the anterior/posterior and medial/lateral will increase as balance is less controlled.

**Delimitations and Limitations**

**Delimitations**

This study will evaluate typically developing preadolescents between the ages of 5-11 years old. The subjects cannot be diagnosed with any mental or physical condition that would delay development, such as Down Syndrome, Autism spectrum disorder, intellectual disabilities, cerebral palsy, etc.

Potential participants will be excluded for any of the following reasons:

- Inability to hop on one leg, unperturbed, for a span of 20 seconds.
- Preexisting injuries or conditions that would prevent them from performing multiple hopping trials of 20 seconds each.
- They do not have the mental capacity to follow directions and follow the beat of a metronome.
- Uncorrected abnormal vision or hearing.
- Cardiac, metabolic or respiratory conditions that could be made worse with physical activity.
- History of seizures.
- Current medications that would affect movement, mental capacity or concentration.

**Limitations**
The subjects will be performing a very specific movement, hopping on one leg with their hands on their hips. The results will provide insight into this population’s motor control strategies, especially in regards to other rebounding movements that follow a mass-spring model such as running, galloping, or bounding. However, the specificity of this movement will limit the ability to generalize the results to other movements. For example, in daily activities an individual is not limited to the use of only one limb. The individual will potentially have the use of the upper extremities, as well as the other leg.

The metronome used to coordinate hopping frequency could also potentially limit the results of the study due to the necessity of sensorimotor integration. A study by Moran et al. (2013) evaluated young adults diagnosed with Expressive Language Impairments Autism Spectrum Disorder (ELI-ASD) hopping at different frequencies (Moran et al., 2013). The subjects were asked to hop on two legs at a self-selected cadence (control = 2.54 ± 0.49 hops/sec and ELI-ASD = 2.21 ± 0.44 hops/sec) and then to the beat of a metronome at 2.3 Hz and 3.0 Hz. At 3.0 Hz the subjects with ELI-ASD were unable to effectively stiffen their legs and match the higher frequency. It was concluded that the sensorimotor deficit characteristic of this population prevented them from matching their movements to the beat of the metronome when it became too fast (Moran et al., 2013). The auditory cue of the metronome might provide an extra challenge to the subjects due to the need to coordinate sensory information with movement.

Definitions

- **Aerial phase**: the time of the hopping cycle when the foot is not in contact with the ground.
- **Autism spectrum disorder**: a range of conditions whose symptoms may include but are not limited to communication problems, social difficulties, motor deficits, decrease in sensorimotor integration, and delayed cognitive functioning.

- **Chronological age**: the age of the individual denoted by the years they have lived. This age can differ from physical or mental age.

- **Degrees of freedom**: the number of directions or coordinates that completely describe a system and movement.

- **Effective leg stiffness**: the stiffening of the musculoskeletal system during stance phase; increases in leg stiffness increases hopping height and frequency and decreases contact time.

- **Flight Phase**: see Aerial Phase.

- **Fourier Transform**: mathematical transformation of time unit into a unit of frequency.

- **Geometry**: the orientation of the limb and the angles of the joints.

- **GEV**: joint variations that will result in a stable end goal task value.

- **Hopping**: unperturbed rebounding vertically off of a single foot; mimics a mass-spring system.

- **Horizontal displacement**: the movement of the subject in the anterior/posterior and medial/lateral direction.

- **Individual variation control strategy**: a strategy where the nervous system will change the stiffness of only one joint, typically the most important joint of the movement, in order to change the system.

- **Intellectual disabilities**: delayed or lack of cognitive development and abilities.
• **Inter-joint control strategy**: a strategy where the nervous system uses one joint of a system to compensate for the weakness of another joint in that system (also referred to as covariation control strategy).

• **Jacobian matrix**: used in UCM analysis to relate deviations in task-related variables to changes in segment angles.

• **Kinematic**: referring to information of the system in space such as speed, vertical and horizontal displacement, etc.

• **Kinetic**: referring to the information of the forces in the system such as the ground reaction force.

• **Leg length**: the effective length of the leg measured from toe to ASIS marker; increases with extension the ankle, knee, and hip joint.

• **Leg orientation**: the geometric shape of the leg in space.

• **Mass-spring model**: the mechanism used to describe running, hopping, trotting, and galloping. The musculoskeletal system mimics a spring by storing and releasing elastic energy upon impact and while rebounding. The musculoskeletal system achieves this by muscle activation and stretching the tendons.

• **Motor control strategy**: any strategy that is utilized by the nervous system in order to coordinate movement or skill performance.

• **Musculoskeletal system**: refers to the bone, ligaments, tendons, and muscles of a particular system or limb.

• **NGEV**: joint variations that change the end goal task value.

• **Pre-adolescent**: the period of human development just preceding adolescence; in this study, 5-11 years of age.
• **Preferred hopping frequency**: the frequency in which the mass-spring mechanism is maximized, energy cost is minimized, and the movement is most efficient. The accepted cadence that is referred to as the preferred frequency is within the range of 2.0-2.2 hops per second.

• **Redundancy**: use of interjoint compensation by increasing the torque of one joint to make up for another weakened or injured joint to result in the same movement; an infinite number of joint torque combinations that produce the same force.

• **Resonant period of vibration**: period of time in which the system stores and transfers energy from one form to the other. In this case the transfer from elastic to kinetic energy during hopping.

• **Sensorimotor integration**: coordination of sensory information, proprioceptive, somatosensory, auditory, or visual, with production of a movement pattern.

• **Stance phase**: the time during the hopping cycle in which the foot is in contact with the ground.

• **Torsional stiffness**: is used to describe the relationship between angular displacement of a joint and the moment about that joint.

• **Typically developing population**: subjects that will be used as a control. These subjects do not have any diagnosed mental, physical, or developmental conditions.

• **Uncontrolled manifold**: joint variations that do not affect the end value of the movement and are unnecessary to control by the nervous system.

**Literature Review**

Hopping
Hopping, continuously rebounding off the floor on one leg, can be described as a mass-spring model (C. T. Farley et al., 1991). The mass-spring model describes the relationship of the displacement of the center of mass and the storage and release of elastic energy (Figure 1). The mass-spring model has been shown to be an accurate predictor of hopping kinematics, such as time in contact with ground and hopping frequency (C. T. Farley et al., 1991). These accurate predictions are possible because of the relationships between leg stiffness, force and displacement (Austin et al., 2002). The stiffness of the leg can be calculated by dividing the peak ground reaction force during stance phase by the displacement of center of mass, in the following equation, \( k = \frac{F_{\text{peak}}}{\Delta L} \) (Claire T. Farley et al., 1998). This formula is derived from Hooke’s law of springs, which states that force (\( F \)) is a product of the stiffness factor (\( k \)) of the spring and the spring’s displacement (\( d \)), or the equation, \( F = kd \). Taking into account that springs oscillate, the period of time spent in contact with the ground can be calculated using frequency (\( \omega \)) (C. T. Farley et al., 1991; Moran et al., 2013). The resonant frequency of the hopping is the angular velocity of the oscillations and can be calculated as \( \omega = \frac{2\pi}{T} \), where \( T/2 \) is the time of contact with the ground when the vertical ground reaction force is greater than bodyweight. Hopping cycle frequency can then be calculated from the time of contact with the ground because of the mass-spring relationship. A full hopping cycle is described as landing to the successive landing on the same foot. Hopping frequency and time on the ground share an inverse relationship because as the frequency increases the subject is required to spend less time in stance phase per hop.
Figure 1: Illustration of mass-spring model from Farley et al. (1998). $\Delta L$ denotes the displacement of mass during the movement (Claire T. Farley et al., 1998).

Movements that do not mimic a mass-spring system will not have the same relationships between leg stiffness, force and displacement. For a mass-spring system, force is dependent on the displacement, which for hopping is the displacement of the center of mass (C. T. Farley et al., 1991). In a mass-spring system, as the displacement of center of mass increases, force also increases. This dependent relationship is illustrated in Figure 2 by hopping frequencies at and above the preferred frequency. Figure 3 also demonstrates hopping frequencies below the preferred frequency that no longer follow a mass-spring system. In these trials the vertical displacement continues to increase but the vertical force decreases (C. T. Farley et al., 1991). The relationships between leg stiffness, force and displacement makes the mass-spring model easy to evaluate and gives insight about the kinematic variables of the movement.
Figure 2: Graphs from Farley et al. (1991) of four different subjects hopping in place at six different frequencies. A) are trials completed with subjects hopping at normal heights and B) are trials where the subjects hopped at a maximal height (C. T. Farley et al., 1991). These graphs show the relationship of force and center of mass displacement. As COM displacement increases, force continues to increase at frequencies at or above the preferred frequency (P.F. or an arrow) illustrating the mass-spring model. Frequencies below the preferred frequency
decrease in force as COM displacement continues to increase, no longer adhering to the mass-spring model.

A mechanically perfect passive spring will only be able to use the energy stored during compression to rebound the center of mass. The leg, acting as a spring, has the ability to put energy into the system by stiffening the leg through the coordination of muscles, tendons, and ligaments (C. T. Farley et al., 1991). The stiffening of the body segment (in this case the leg) during the stance phase of hopping is known as leg stiffness (C. T. Farley et al., 1991). Hopping is a multi-joint movement requiring the ankle, knee, and hip joint to be coordinated in geometry and torsional stiffness to effectively stiffen the leg (Claire T. Farley et al., 1998). The torsional stiffness of a joint is described as the relationship between angular displacement and the moment about the joint (Claire T. Farley et al., 1998). A stiffer joint for a given moment will result in less angular displacement. Geometry of the leg can also affect leg stiffness by changing the joint angles and the muscle-tendon lengths of the leg (Claire T. Farley et al., 1998). A greater joint moment will occur when the joints are more flexed at contact, thereby reducing the effectiveness of the leg stiffness. Also the geometry of the leg will change the muscle-tendon lengths, changing the required muscle activation to create a sufficient force to rebound off the ground (Claire T. Farley et al., 1998).

A study by Farley and Morgenroth (1999) observed the ankle joint as the primary joint that affects leg stiffness during hopping. Their subjects were instructed to hop as high as possible while maintaining a certain frequency. The subjects would need to increase their leg stiffness in order to increase their hopping height. The frequency was required to stay the same as a controlled variable because frequency will also affect the effective leg stiffness. This effect
will be discussed later on. Compared to preferred hopping height, the subjects increased their ankle stiffness 1.9-fold and their knee stiffness 1.7-fold when hopping for maximum height. After running these results in a computer simulation model of hopping, leg stiffness increased 2.0-fold as a result to changes at the ankle. The changes at the knee did not affect leg stiffness. These results indicated that the ankle is the primary joint that is stiffened in order to increase leg stiffness (Claire T. Farley & Morgenroth, 1999). The stiffened knee joint may have increased in order to change the geometry of the leg. It has been found that if an individual adds an extended knee at mid-stance then they are able to double the effect of a stiffened ankle (Claire T. Farley et al., 1998).

Another study by Chang et al. (2008) looked at how the leg would compensate if one resisted the ankles’ ability to effectively stiffen. Their subjects wore an ankle foot orthosis (AFO) that resisted plantarflexion of the ankle. They were then required to hop at frequencies greater than the preferred 2.2 hops/second. These frequencies would require the subjects to increase the stiffness of their leg in order to achieve performance success. They found that when plantarflexion was inhibited the subjects used a multi-joint compensation strategy by stiffening both the ankle and the knee joints. Contrary to previous studies, the subjects increased knee flexion to compensate with plantarflexion resistance. Knee flexion during hopping would typically decrease leg stiffness. A potential reason for the use of this strategy may be to take advantage of their knee extensor muscles (the quadriceps) to create more power in the ankle joint. Or perhaps the flexed knee position helps activate a neural mechanism that increases ankle extension. This study demonstrated that the subjects employed a multi-joint compensation strategy when the ankle’s ability to plantarflex was impaired (Chang et al., 2008).
An important question when dealing with motor control is how the body deals with the degrees of freedom associated with movement (Scholz & Schöner, 1999). The first theory came from Bernstein (1967), which stated the nervous system controlled the spatial aspects of the movement rather than the specific joints involved. Since this first theory there have been many different perspectives on the problem of degrees of freedom (Scholz & Schöner, 1999). These different perspectives have included control of the trajectories of the limb in space, the limb movement being coordinated by joint coordinates, rotation about specific joint axes, and the use of center of mass information. More recently, another method has been developed to look at the degree of freedom problem by studying the structure of variability of movement (Scholz & Schöner, 1999).

The uncontrolled manifold (UCM) analysis looks at two different spaces for movement: the joint configurations that will lead to the same value of the controlled variables and the joint configurations that change the controlled variables’ values (Scholz & Schöner, 1999). The uncontrolled manifold is the joint combinations that do not affect the end variable of the movement, therefore they are unnecessary for the nervous system to control. In order to quantify the UCM it has to be linearized using the mean joint configuration from the end position. This linearization requires the assumptions that the joint configurations can be compared to each other and that the movement is coordinated by a sequence of postural states. Essentially, uncontrolled manifold allows a researcher to find how variability of joint motion is partitioned between the uncontrolled subspace and its perpendicular subspace. A UCM analysis will also allow the researcher to find the timing of each variable control throughout the movement (Scholz & Schöner, 1999)
This theory of controlling human movement also infers that there may be many different joint solutions to achieve the same end movement goal (Tseng, Scholz, & Schöner, 2002). When using UCM there are two different components that are assessed. The component that includes the variations that change the variable task value is called non-goal-equivalent variability (NGEV) (Tseng et al., 2002). These variations are mathematically described as being perpendicular to the linearized UCM (Figure 3) (An, Asama, Stepp, & Matsuoka, 2011). The other component is called goal-equivalent variability (GEV) and includes those variations that result in a stable value of the variable task (Tseng et al., 2002). These variations are parallel to the linearized UCM model (An et al., 2011). Comparing these two values allows the researcher to decide if the nervous system is utilizing a control on that variation (Tseng et al., 2002). If the GEV is larger than the NGEV then there is a range of solutions being used to reach task success and control the variables associated with the movement. If the NGEV is larger than the GEV then task success is not dependent on control of the variables being evaluated. If both GEV and NGEV are close to zero then the nervous system is using more resources and tightly controlling that variation because a unique joint coordination is needed to achieve the goal task (Tseng et al., 2002). The variability of a movement pattern may also describe a highly variable system due to lack of efficient technique (Black, Smith, Wu, & Ulrich, 2007). For example, a novice athlete will be more varied in their movement to achieve a sport-specific task, while an elite athlete will be more efficient (Black et al., 2007).
Figure 3: An example of a UCM analysis taken from An et al. (2011). The solid curved line is the linearized UCM. The white circle indicates the average joint angle at that postural state. Using the line and the average joint angle data point, both the parallel null space \( \mathcal{E}_\parallel \) and the perpendicular null space \( \mathcal{E}_\perp \) can be found. These two vectors are then used as references to determine the ratio of each data point collected (An et al., 2011).

**UCM and Hopping**

Since the introduction of UCM analysis, the majority of the studies have focused on the upper extremity including posture and pointing (Auyang et al., 2009). A study by Auyang, Yen and Chang (2009) used UCM analysis to study hopping motor control for young adults. They looked at the control of both the leg length and leg orientation throughout a hopping cycle. The leg vector was defined from the anterior superior iliac spine marker to the toe marker. Pelvis, thigh, shank, and foot segment angles were defined along this vector and angular displacement of each segment was calculated from initial contact to mid-stance. The leg length mathematical model was defined as the differences of leg vector magnitude in relation to segment angle changes. Similarly, the leg orientation mathematical model used changes in segment angle but to
describe changes in leg orientation. The range of leg orientation was found by the difference of the maximum and minimum angles of the leg vector in relation to the ground during stance phase. The results showed that the subjects did not control their leg length and leg orientation throughout the entire cycle. Instead they found that leg length was controlled at mid-stance and leg orientation during the aerial phase (Figure 4). The need to control the leg length at mid-stance could be caused by the need to effectively stiffen the leg (Auyang et al., 2009). As stated before, the joint torsional stiffness and the knee being extended or flexed can influence the overall stiffness of the leg. So their results would indicate that the subjects were attempting to control the angular displacement and geometry of the leg at mid-stance in order to effectively stiffen their leg. Control at mid-stance might also be needed to effectively change the joints from flexing to extending (Auyang et al., 2009). The nervous system controlling how the body lands and the movement of the center of mass can be explained by the stabilization of leg orientation during the aerial phase. An interesting note from this study is that while one of these variables was maximally controlled, the other was minimally controlled. This trade-off gives insight into how the nervous system will economize the cost of movement (Auyang et al., 2009).

Figure 4: Charts from Auyang et al. (2009) of the Index of Motor Abundance for both leg length (a) and leg orientation (d) for preferred 2.2 hops/second. The Index of Motor Abundance is the
amount of variability in the uncontrolled subspace. A significant positive IMA illustrates variable control for task performance. Leg length IMA is significantly positive during mid-stance and decreases during aerial. Leg orientation IMA is significantly positive during aerial phase and decreases during contact (Auyang et al., 2009).

The tradeoff between stabilizing leg length and leg orientation during hopping is also found when a subject attempts to hop in a confined space (Auyang & Chang, 2013). Auyang and Chang (2013) asked their subjects to hop within three different target areas that were increasingly smaller. They then used UCM to quantify the control strategies of leg length and leg orientation and how the strategies might change when the subjects were confined to smaller areas. They found that when the target space became smaller the subjects’ control of leg length decreased and the control of leg orientation increased. This finding demonstrates the nervous system’s strategy to only control aspects of the movement that will correspond with performance (Auyang & Chang, 2013). Leg orientation for this task is more important because the subjects were asked to control their foot placement during hopping. By decreasing the control of leg length, the nervous system was allowing more joint angle variance combinations available to the subject so that the leg orientation was not affected (Auyang & Chang, 2013). However, the nervous system does have the ability to tightly control both factors when it is necessary for performance. At mid-stance leg length was tightly controlled even during the smaller target trials when leg orientation was the main factor influencing success (Auyang & Chang, 2013). As stated before, mid-stance is an important phase in the hopping cycle for leg length because of the change from flexing to extending and the need to stiffen the leg in order to rebound off the floor (Auyang et al., 2009). The ability of the nervous system to control both leg length and leg
orientation shows that there is plenty of space for the control of both factors. However, it seems that the nervous system will only control factors at crucial times when they might affect performance (Auyang & Chang, 2013). One interesting note is that the actual foot placement performance did not increase as the target areas became smaller (Auyang & Chang, 2013).

Another study by Yen, Auyang, and Chang (2009) looked at the potential for a kinetic redundancy strategy during hopping in place for both vertical and horizontal forces. They used UCM and looked at the NGEV and GEV of joint torque variance during beginning (landing), middle (mid-stance), and end (take-off) of stance phase. High GEV compared to NGEV would indicate stabilization during that section of stance phase. The vertical force was stabilized during landing, mid-stance, and take-off and the horizontal force was destabilized during landing and take-off. The control of the vertical force during stance phase is important to limit the amount of energy lost at contact and to produce forces to hop at a consistent frequency. Stabilization of vertical force during landing, mid-stance, and take-off would also result in consistent kinematic variables such as stance phase time and hopping height. The destabilization of horizontal forces at landing and take-off allows the subject to correct any horizontal displacement that is occurring during hopping (Yen, Auyang, & Chang, 2009).

**Frequency as a Variable**

A method to make the simple task of hopping more difficult is to increase the frequency the subject must hop (Auyang et al., 2009). There is a resonant hopping frequency and when subjects are forced to hop outside of it more muscle force is required, more energy is expended and their movement is less efficient (Auyang et al., 2009). The preferred frequency, or frequency that best resembles the mass-spring system, is within the range of 2.0-2.2 hops per second in healthy young adults (Austin et al., 2002; C. T. Farley et al., 1991; Ferris & Farley,
A study by Austin, Tiberio and Garrett (2002) examined if the leg would still act as a mass-spring mechanism if the frequency of the hopping deviates from the preferred hopping rate. They compared the vertical stiffness and angular displacements of the ankle, knee and hip joints when the subjects hopped at a preferred frequency (2.03Hz), and plus or minus 20% of that frequency. They found at preferred and +20% frequency a linear best fit line describing the relationship of force and displacement of the center of mass. This would illustrate a simple mass-spring relationship. However, when the frequency was -20% they did not find a linear correlation and therefore a mass-spring relationship was not present. They also found an expected change in angular displacements with respect to frequency. At an increased frequency there was a decrease in flexion at the three joints. At a slower frequency there was an increase in flexion at each joint (Austin et al., 2002). These results match previous research and demonstrate the changes needed for an individual to hop at frequencies other than the resonant frequency.

Changing the frequency of hopping will affect the mechanics more than adding mass to the system. Another study by Austin, Tiberio, and Garrett (2003) looked at the effects on vertical stiffness and angular displacement of the ankle, knee and hip joints when the subjects hopped at different frequencies and mass was added to the system. The three frequencies they used were preferred, +20% and -20% and they used three different mass models: unweighted, +10% body mass and +20% body mass. Their results indicated significant differences in vertical stiffness and ankle dorsiflexion as a variation of frequency, but not mass (Austin et al., 2003). The significant difference of ankle dorsiflexion is important because the ankle has been shown to be the most influential joint on leg stiffness (Claire T. Farley & Morgenroth, 1999). This study
demonstrates that changing the frequency of the hopping is effective enough to load the leg and result in a change of mechanics (Austin et al., 2003).

A study by Yen and Chang (2010) used UCM analysis to look for a potential change in control strategy when subjects were asked to hop at faster frequencies. They looked at two potential strategies, covariation of the joints and individual variation. Covariation of the joints would be present if the subjects changed the torque of one joint to compensate for another joint (Yen & Chang, 2010). For example, this was demonstrated by the previously discussed study (Chang et al. 2008) where the knee joint compensated for the ankle during impaired plantarflexion. The individual variation strategy would be present if the subjects stabilized the most important joint for the movement, which for hopping is the ankle joint (Yen & Chang, 2010). At the preferred frequency the subjects used a covariation strategy to fix joint fluctuations between hops. As they increased the frequency, the joint variances decreased with the ankle decreasing the most. The decrease of the ankle joint variance had the most weight on the stance phase time and cadence. These results indicated the subjects were utilizing an individual variation strategy. Essentially, the subjects used two different control strategies based on the frequency that they were hopping. Increasing the hopping frequency requires a switch from covariation to an individual variation strategy by decreasing the variance of the ankle joint (Yen & Chang, 2010).

Not only does increasing the frequency of hopping require a change of motor control strategy, but it also requires more stability control (Auyang et al., 2009). The Auyang et al. (2009) study that evaluated the control of leg length and leg orientation throughout the hopping cycle also compared frequencies. They collected data of their subjects hopping at 2.2, 2.8 and 3.2 hops/second. They then compared the Index of Motor Abundance (IMA) for leg length and
leg orientation across all frequencies (Figure 5). Their results illustrated that the IMA for leg length significantly increased with frequency, demonstrating an increase in stability control. Leg orientation did not increase with frequency because task performance was not dependent on this variable (Auyang et al., 2009). Frequency, as a variable, increases the difficulty of hopping and affects the mechanics of control and stability.

Figure 5: Charts from Auyang et al. (2009) illustrating the average IMA for leg length (a) and leg orientation (b) across all frequencies. IMA of leg length at 2.8Hz is significantly different from 2.2Hz. IMA of leg length at 3.2Hz is significantly different from both 2.2 and 2.8Hz (Auyang et al., 2009).

Pre-adolescent Hopping

All of the studies discussed so far on the mechanics and motor control of hopping have been done with typically developing adults. A study by Lang, Busche, Rakhimi, Rawer and Martin (2013) used subjects aged 3-19 years old. This age group would include children still developing and others who have fully developed. In this study they looked at the maximum voluntary force and the whole body stiffness of their subjects when hopping on one leg. They normalized their results to bodyweight because of the age disparity. They found that their
subjects between the ages of 3-5 could not coordinate the single-leg hopping and produced less than expected peak forces. After age 5 a constant mass relative maximum voluntary force relationship was found with age, regardless of gender. However, there was a side difference between dominant and non-dominant legs of 5.6% lower peak force in the non-dominant leg. Peak whole body stiffness was dependent on the age of the subject. From the ages 7-19 there was a linear increase in whole body stiffness in regards to age. Gender did not play a role in whole body stiffness until the ages of 15-19. At these ages males continued to increase while the females stayed constant (Lang et al., 2013). When whole body stiffness was normalized to body mass there was a decrease in standard deviation as age increased. This study demonstrates that there is a developmental learning curve for single leg hopping until the age of six. After the age of six an individual’s ability to effectively stiffen their leg increases with age (Lang et al., 2013). These results would indicate the need to study motor control strategies of a still developing, pre-adolescent age group.

Studying the control strategies of a pre-adolescent age group will provide a better understanding of the development of the motor control for mass-spring model movements, such as running, bounding, and hopping. These movements are used every day by children for play, sports, and daily activities. The ability of the uncontrolled manifold analysis to quantify the stability of control allows the researcher to understand and compare the movement efficiency of the population. Further research will then be able to compare other populations, such as special populations whose development and motor control is typically delayed. Down syndrome is a special population that has well-documented movement variability, cognitive delays, and motor control deficits (Fidler, Hepburn, Mankin, & Rogers, 2005; Frith & Frith, 1974; Jobling, 1994; Roizen, 2001; Shumway-Cook & Woollacott, 1985). A study by Black et al. (2007) used UCM
analysis to evaluate and compare treadmill-walking variability of typically developing pre-adolescents and children with Down syndrome. Surprisingly, their results indicated that the children with Down syndrome partitioned more goal equivalent variance than the typically developing population. This would usually indicate more efficient movement and control; however, previous research and observation of the movement would not indicate the Down syndrome population as better movers. They concluded that the Down syndrome children were actually partitioning more goal equivalent variance because their movement is highly variable. This would allow them a greater range of solutions to complete the task and therefore allow them to recover from their inefficient movement patterns (Black et al., 2007). Results like these indicate the need for a better understanding of how motor control strategies develop and the need for typically developing control data. Studying the typically developing pre-adolescent population would provide both the comparison data for special populations and an insight into the development of mass-spring motor control strategies compared to an adult population.

**Methods**

**Recruitment and Data Collection at the Lab**

We recruited 15 subjects consisting of 8 males and 7 females between the ages of 5-11 years old. Of the 15 recruited subjects, 13 of the subjects responded to the different frequency conditions. Two of the subjects, aged 6 and 7, did not change their hopping frequency potentially due to lack of understanding, the lack of desire to cooperate, or the inability to change frequency. The remaining 13 subjects were used for data analysis. Table 1 contains anthropometric measures of the subjects analyzed. The data collection was held in the biomechanics lab at Georgia State University. Upon arrival the child was given a period of time
to acclimate to the lab environment by walking around or sitting in a chair. While the child became familiar with the environment a verbal overview of the process was given to the parents and the child. At the end of the overview the parents were asked to sign an Informed Consent form for their child (Appendix A). A script was read to the child for his/her verbal consent. (Appendix B).

Table 1: Anthropometric measures of subjects analyzed. Subjects who did not complete condition MF are signified by DNC.

<table>
<thead>
<tr>
<th>Subject ID</th>
<th>Age</th>
<th>Gender</th>
<th>Height (cm)</th>
<th>Mass (kg)</th>
<th>Leg Length (cm)</th>
<th>Hopping Foot</th>
<th>MF Condition</th>
</tr>
</thead>
<tbody>
<tr>
<td>S02</td>
<td>9</td>
<td>Female</td>
<td>136</td>
<td>33.18</td>
<td>72.75</td>
<td>R</td>
<td>Completed</td>
</tr>
<tr>
<td>S03</td>
<td>9</td>
<td>Female</td>
<td>129</td>
<td>30.45</td>
<td>67.75</td>
<td>R</td>
<td>Completed</td>
</tr>
<tr>
<td>S04</td>
<td>5</td>
<td>Male</td>
<td>122</td>
<td>25.00</td>
<td>64.00</td>
<td>R</td>
<td>DNC</td>
</tr>
<tr>
<td>S05</td>
<td>8</td>
<td>Male</td>
<td>131</td>
<td>30.45</td>
<td>67.25</td>
<td>R</td>
<td>DNC</td>
</tr>
<tr>
<td>S06</td>
<td>6</td>
<td>Male</td>
<td>124</td>
<td>26.36</td>
<td>65.25</td>
<td>R</td>
<td>Completed</td>
</tr>
<tr>
<td>S07</td>
<td>10</td>
<td>Female</td>
<td>127.5</td>
<td>31.82</td>
<td>67.60</td>
<td>R</td>
<td>Completed</td>
</tr>
<tr>
<td>S08</td>
<td>11</td>
<td>Male</td>
<td>137</td>
<td>33.64</td>
<td>74.65</td>
<td>R</td>
<td>Completed</td>
</tr>
<tr>
<td>S09</td>
<td>10</td>
<td>Male</td>
<td>141</td>
<td>34.09</td>
<td>74.50</td>
<td>L</td>
<td>Completed</td>
</tr>
<tr>
<td>S10</td>
<td>8</td>
<td>Male</td>
<td>130</td>
<td>24.55</td>
<td>64.60</td>
<td>R</td>
<td>DNC</td>
</tr>
<tr>
<td>S11</td>
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<td>Female</td>
<td>150</td>
<td>45.45</td>
<td>81.75</td>
<td>L</td>
<td>Completed</td>
</tr>
<tr>
<td>S12</td>
<td>8</td>
<td>Female</td>
<td>124.5</td>
<td>27.27</td>
<td>66.25</td>
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<td>Completed</td>
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<tr>
<td>S14</td>
<td>10</td>
<td>Male</td>
<td>143.5</td>
<td>53.18</td>
<td>76.25</td>
<td>L</td>
<td>Completed</td>
</tr>
<tr>
<td>S15</td>
<td>10</td>
<td>Male</td>
<td>137</td>
<td>29.09</td>
<td>73.65</td>
<td>R</td>
<td>Completed</td>
</tr>
<tr>
<td>Mean</td>
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<td>133.27</td>
<td>32.66</td>
<td>70.48</td>
<td></td>
<td></td>
</tr>
<tr>
<td>SD</td>
<td>1.75</td>
<td></td>
<td>8.02</td>
<td>7.87</td>
<td>5.24</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

A full body PSIS marker set was marked onto the subject with a washable eyebrow pencil. The following locations were marked on both the right and left sides of the body: toe, ankle, heel, shank, knee, thigh, anterior superior iliac spine, posterior superior iliac spine, elbow, wrist, front of head and back of head. Sternum, clavicle, C-7, T-10, and shoulder were marked on the trunk. Anthropometric measurements were then taken using the marks including foot
width, ankle width, knee width, elbow width, wrist width, hand thickness, inter-ASIS distance, leg length, and shoulder offset. Once the measurements were taken, reflective markers were attached at the marks and reinforced with tape.

The subject was then instructed to stand on the force plate and hold a T-pose for calibration. For the T-pose, the subject stood upright and held their arms out at 90 degrees from the body. Five seconds of data were captured while holding the T-pose. The subject was then given verbal and visual instructions of single leg hopping on the force plate. They were instructed to hop on their dominant leg, the leg they would kick a ball with, and place their hands on their hips. They were given the goals to hop continuously and to stay on the force plate until given the instruction to stop. The subject was then given adequate time to practice single leg hopping. After the subject felt comfortable single leg hopping, 3 trials were collected at their preferred self-selected speed. These trials were 20 seconds in duration. A successful trial consisted of the subject hopping continuously for 20 seconds, not hopping off of the force plate, and keeping their hands on their hips.

After three successful trials were collected, at their preferred frequency, the average frequency was calculated. Next the subject was introduced to the metronome that set the pace for the following conditions. The metronome provided an auditory cue of the frequency that the subject should attempt to hop. Once the subject was familiar with the metronome, four more conditions were collected. The metronome was set to one of four frequency conditions, which were randomized across all subjects to reduce the potential for a learning effect. The four frequency conditions consisted of the individual’s calculated average preferred frequency (MP), a 20% increase (MM) of MP, a 40% increase (MF) of MP, and a 20% decrease (MS) of MP. The subject was asked to match the frequency to the best of their ability. For these conditions a
successful trial consisted of the same criteria used for the free hopping trials along with visually matching the beat of the metronome. The subjects were verbally encouraged to match the metronome throughout the trial if they were not visually keeping pace. Three successful trials were collected at each frequency condition. Three subjects were unable to complete the MF condition because their preferred frequency was elevated and the MF frequency was physically impossible. Plenty of rest was provided between each trial and condition in order to reduce a fatigue effect.

Safety

The only safety concern during the study was the potential injury to the subject while hopping on one foot. A force plate is not a compliant surface and so continuously hopping on the same foot multiple times could cause stress. However, the potential for real injury during this study was assumed to be low compared to the jumping activities children complete in daily life. A research assistant was standing by in case the participant lost his/her balance during hopping.

Data Recording and Analysis

The kinematic data were recorded using Vicon Nexus (Oxford, UK) and seven cameras with a sampling rate of 100 Hz. Kinematic data were analyzed using a coordinate system in which movement in the upward direction was positive and movement in the downward direction was negative. In a similar sense, movement in the horizontal could either be negative or positive depending on direction. In reference to subject orientation, forward was positive and backward was negative in the AP direction, and to the right was positive and to the left was negative in the ML direction. Absolute values were used during anterior/posterior and medial/lateral average calculations. Kinetic data of ground reaction force were collected using a force plate with a
sampling rate of 1000 Hz (AMTI, MA, USA) located in the floor of the lab. For each trial, hops that met the data collection criteria were used to calculate trial means. Condition means for each subject were then calculated using the trial means.

**Hopping Frequency**

In order to measure deviation of actual hopping cadence from the metronome frequency a Fourier Transform was used on the vertical ground reaction force data (Moran et al., 2013). The Fourier Transform transforms the time unit into the hopping frequency that was prevalent throughout the trial (Figure 6).

![Figure 6: Representative vertical GRF data from a subject (A) that has had a Fourier Transform applied in order to produce the hopping cadence (B).](image)

The following formula was then used to quantify the percent deviation \(d\) between the cued frequency \(\omega_{\text{cue}}\) and the actual frequency \(\omega_{\text{actual}}\) (Moran et al., 2013).

\[
d = \left(\frac{\omega_{\text{cue}} - \omega_{\text{actual}}}{\omega_{\text{cue}}}\right) \times 100\%
\]  

(1)

**Leg Stiffness**

The effective leg stiffness of the subject was calculated using the linear regression slope of the ground reaction force vs. center of mass displacement over the entire stance phase of a hop.
The ground reaction force was normalized to bodyweight and then plotted with the displacement of the center of mass (COM). This relationship is indicative of the mass-spring model and is described by the following formula derived from Hook’s Law.

\[ k = \frac{F}{\Delta L} \]  

where \( k \) is stiffness, \( F \) is vertical ground reaction force, and \( \Delta L \) is the displacement of the COM. Two different methods were used to calculate the displacement of the center of mass. The first method used marker data to calculate center of mass displacement. The Full Body Plug-In Gait Model calculated a center of mass data point during each trial based on marker and anthropometric measures. Displacement of the center of mass was then calculated using the position of the center of mass at foot strike (\( p_0 \)). The displacement of the center of mass at each time index was found by subtracting the position from \( p_0 \). This method of calculation is not found in the literature, however, it is referenced as a possible method to calculate COM displacement and leg stiffness (Butler et al., 2003).

The second method has been extensively used and calculates the double integration of acceleration from the force plate data (Austin et al., 2002, 2003; Chang et al., 2008; C. T. Farley et al., 1991; Claire T. Farley et al., 1998; Claire T. Farley & Morgenroth, 1999; Ferris & Farley, 1997). Double integration of acceleration followed the \( S_p^1 \) method outlined by Hébert-Losier and Anders Eriksson (Hébert-Losier & Eriksson, 2014). Acceleration was found using the following formula,

\[ a_i = \frac{1}{m} f_i - g \]  

where \( a_i \) is acceleration (m/s\(^2\)) at time index \( i \), \( m \) is mass (kg), \( f_i \) is force (N) at time index \( i \), and \( g \) is acceleration due to gravity (9.81 m/s\(^2\)). Velocity was then calculated using the formula,
\[ v_i = v_{i-1} + \Delta \tau a_i \]  

(4)

where \( \tau \) is time index of .001 seconds. Velocity was then integrated into position change with the formula,

\[ p_i = p_{i-1} + \Delta \tau v_{i-1} \]  

(5)

Velocity and position were then adjusted using the ‘P’ method, which operates under the assumption that at takeoff the COM position is considered zero. Therefore, \( v_0 \) was calculated with the following integration constant,

\[ v_0 = -p^*(T)/T \]  

(6)

where \( p^*(T) \) is the unadjusted position at takeoff and \( T \) is time index at takeoff. Velocity and vertical displacement were then adjusted with the new \( V_0 \) integration constant (Hébert-Losier & Eriksson, 2014).

Toe Marker Movement

In the vertical direction, hopping height was obtained for each hop by calculating the range of the toe marker in the vertical direction during each flight phase. The difference between the foot on the ground and the highest position of the toe marker in the vertical direction illustrates hopping height. Peak vertical velocity of the foot was also calculated using the vertical toe marker data. Displacement of the toe marker in the vertical direction during flight phase was calculated and then divided by the time per frame of 0.01 seconds. The maximum negative velocity value was used as the peak vertical velocity because the peak vertical velocity occurred just before landing. The timing of the peak vertical velocity could potentially be relevant to foot injury.
Figure 7: Representative toe marker data in the vertical direction during flight phase of each hop for one trial. A) Graph of the position of the toe marker in the vertical direction over time, from which B) the velocity of the toe marker in the vertical direction can be calculated and plotted.

In the medial/lateral (ML) and anterior/posterior (AP) directions, displacement of the toe marker was calculated by finding the absolute difference of the toe marker at each foot strike. The range of the toe marker was calculated using the maximum position of the toe marker and subtracting the minimum position. The range was further used to find an approximate area that the subject hopped within during the entirety of a trial. An ellipse area formula was used to find the area,

$$A = \pi ab$$

(7)

where $A$ is area, $a$ is $\frac{1}{2}$ the range in the AP direction and $b$ is $\frac{1}{2}$ the range in the ML direction. The combination of the ranges and area demonstrates the amount of space that the subject covered while hopping (Fig. 8).
Figure 8: Plot of the position of the toe marker at each foot strike in reference to the force plate and the first foot strike of the trial. The four corners of the force plate are plotted as triangles. This trial demonstrates good balance control with little displacement between foot strikes and small area covered by the subject.

Center of Mass Movement

Displacement of COM was calculated for each hop during stance phase in the vertical, AP, and ML directions. In the vertical direction the maximum COM displacement was calculated during stance phase for each hop. Vertical COM displacement was calculated using the formula,

\[ Displ = COM_{fs} - COM_{MIN} \]

where \( COM_{fs} \) is the position of the center of mass at foot strike and \( COM_{MIN} \) is the position of the center of mass when it was the lowest. The average velocity of the center of mass during stance phase was calculated for each hop. For each time index, the displacement of the center of mass was calculated from the previous position and divided by 0.01 seconds. The
average of the absolute values of these velocities was then calculated to illustrate how quickly the subject changed position during the stance phase of hopping. Also the peak vertical velocity of the center of mass in the flight phase was calculated. This velocity was calculated using the same method as the toe peak velocity in that the maximum negative value was used.

Range of COM, in the horizontal, was calculated as maximum position minus the minimum position of the COM data point generated by the Full Body Plug-In Gait Model. The range of the COM illustrates the displacement of the COM during stance phase for each hop and highlights the amount of postural sway. Smaller ranges correlate to small displacements of COM around a central position. COM velocity in the ML and AP direction was also calculated to look at balance control during stance phase. Finding the absolute difference between COM positions at each time index and dividing by 0.01 seconds calculated COM velocity.

**Statistical Analysis**

Our hypothesis is that typically developing preadolescents will coordinate hopping at different frequencies similarly to young adults. The preadolescent subjects will increase leg stiffness with an increase in frequency and decrease leg stiffness with a decrease in frequency. As frequency increases there will be more toe marker movement in the AP and ML directions as movement in the horizontal becomes less important to task success. Similarly, as frequency increases there will be more COM movement in the AP and ML directions as it becomes less important to match the higher frequencies.

A one-way (frequency) ANOVA test with repeated measures was conducted on each dependent variable. The dependent variables for leg stiffness included leg stiffness using the marker method and leg stiffness using the force plate method. The dependent variables
evaluating toe marker movement were AP/ML movement of the toe marker, hopping height, peak vertical velocity of the toe, AP/ML range of the toe marker, and hopping area. The dependent variables analyzing COM movement were maximum vertical displacement of the center of mass, average vertical velocity of the center of mass, and the peak vertical velocity of the center of mass, AP/ML range of motion of the center of mass and AP/ML velocity of the center of mass. The within-subject grouping factors were the frequencies including metronome guided preferred (MP), a 20% increase (MM), a 20% decrease (MS) and a 40% increase (MF). Significant differences were concluded at an alpha level of $\alpha = 0.05$. In the case of significant differences, a Least Significant Difference pair-wise post-hoc analysis was completed to find where the differences between groups were present.

Results

Subjects

The average age, height, weight and leg length of the subjects were 8.85 years old, 133.27cm, 32.66kg, and 70.48cm, respectively (Table 2). There were no significant differences in anthropometric measures between males and females.

Table 2: Mean (SD) subject anthropometric information.

<table>
<thead>
<tr>
<th>Gender</th>
<th>Count</th>
<th>Age</th>
<th>Height (cm)</th>
<th>Weight (kg)</th>
<th>Length (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Male</td>
<td>8</td>
<td>8.50(2.14)</td>
<td>133.19(7.75)</td>
<td>32.05(9.28)</td>
<td>70.02(5.20)</td>
</tr>
<tr>
<td>Female</td>
<td>5</td>
<td>9.40(1.14)</td>
<td>133.40(10.19)</td>
<td>33.64(6.96)</td>
<td>71.22(6.38)</td>
</tr>
<tr>
<td>Group</td>
<td>13</td>
<td>8.85(1.82)</td>
<td>133.27(8.35)</td>
<td>32.66(8.19)</td>
<td>70.48(5.45)</td>
</tr>
</tbody>
</table>
Frequency

The mean preferred frequency for subjects was 2.55 Hz. The average maximum percent increase, excluding subjects who did not complete the MF condition, was 28.59% and maximum percent decrease was -14.87%. Deviations from the cued frequency were averaged for each condition and subject (Table 3). The MP, MM, and MF conditions were less than a 10% difference on average and the MS condition was greater than a 10% difference.

Table 3: Frequency accuracy across all subjects for each condition, including the average difference for that condition, standard deviation and the count of subjects for each category.

<table>
<thead>
<tr>
<th>Condition</th>
<th>Actual Freq (Hz)</th>
<th>Cued Freq (Hz)</th>
<th>%diff</th>
<th>5% diff</th>
<th>10% diff</th>
<th>&gt;10%diff</th>
</tr>
</thead>
<tbody>
<tr>
<td>MS</td>
<td>2.25(0.28)</td>
<td>2.12(0.30)</td>
<td>11.33%(9.20%)</td>
<td>5</td>
<td>8</td>
<td></td>
</tr>
<tr>
<td>MP</td>
<td>2.56(0.32)</td>
<td>2.55(0.031)</td>
<td>1.13%(1.35%)</td>
<td>13</td>
<td></td>
<td></td>
</tr>
<tr>
<td>MM</td>
<td>2.96(0.35)</td>
<td>3.05(0.37)</td>
<td>3.65%(2.91%)</td>
<td>8</td>
<td>5</td>
<td></td>
</tr>
<tr>
<td>MF</td>
<td>3.08(0.33)</td>
<td>3.41(0.32)</td>
<td>9.53%(6.68%)</td>
<td>3</td>
<td>2</td>
<td>5</td>
</tr>
</tbody>
</table>

Leg Stiffness

Leg stiffness increased with an increase in frequency. There was a frequency effect on leg stiffness calculated by the markers (F(3,45) = 93.88, p < .001, Fig. 9) and on leg stiffness calculated by the force plate (F(3,45) = 79.93, p < .001, Fig. 10). The markers calculation method resulted in significant increases from MS to MP, MS to MM, and MS to MF. There were also significant increases from MP to MM, MP to MF, and MM to MF (Fig. 9). The force
plate calculation method produced the same significant increases from MS to MP, MS to MM, MS to MF, MP to MM, MP to MF, and MM to MF (Fig.10).

Figure 9: Mean and standard deviation of leg stiffness (BW/m) as calculated from the center of mass marker (* denotes significant difference at p < .05).
Figure 10: Mean leg stiffness (BW/m) and standard deviation for each metronome condition as calculated using double integration of acceleration of the force plate data (* denotes statistical difference at $p < .05$).

**Toe Marker Movement**

In the vertical direction, the subjects decreased hopping height and peak toe velocity before foot strike as frequency increased. Statistical analysis indicates that there was a frequency effect on both hopping height ($F(3,45) = 17.71$, $p < .001$, Fig. 13) and peak toe velocity ($F(3,45) = 10.44$, $p < .001$, Fig. 14). After post-hoc analysis there were significant differences found between frequencies for hopping height between MS to MP, MS to MM, MS to MF, MP to MM and MM to MF. Significant differences between frequencies for peak vertical toe velocity were found between MP to MM and MP to MF.
Figure 13: The mean hopping height (cm) at each condition calculated during the flight phase (* denotes statistical difference at p < .05).

Figure 14: The mean peak vertical velocity (m/s\(^2\)) during flight phase for each metronome condition (* denotes statistical difference at p < .05).
In the horizontal plane the subjects decreased toe displacement in both the AP and ML directions as the frequency of hopping increased. Statistical analysis indicates that there was a frequency effect on the AP toe displacement ($F(3,45) = 19.41$, $p < .001$, Fig. 11a) and ML toe displacement ($F(3,45) = 12.71$, $p < .001$, Fig. 11b). Post-hoc analysis reveals that for AP toe displacement there were significant decreases between all conditions with the trend of decreased displacement with increasing frequency. The ML toe displacement produced significant decreases from MS to MP, MS to MM, MS to MF, MP to MM and MP to MF. No significant effects were found in the AP and ML toe ranges (Fig. 12) and hopping area (Fig. 13).

Figure 11: Mean and standard deviation of toe displacement (cm) at each foot strike A) in the AP direction and B) in the ML direction. Symbol * denotes significant difference at $p < .05$. 

A)  

B)
Figure 12: A) Mean range of the toe marker movement in the anterior/posterior plane. B) Mean range of the toe marker movement in the medial/lateral plane.

Figure 12: The mean hopping area calculated as an ellipse from the range values for each metronome condition.

Center of Mass Movement

Vertical COM displacement, average COM velocity during stance and peak COM velocity during flight decreased with an increase in velocity. There was a frequency effect on maximum vertical displacement of the COM during stance phase (F(3,45) = 51.40, p < .001, Fig. 40
16), average velocity of the COM during stance phase \((F(3,45) = 23.65, p < .001, \text{Fig. 17})\) and the peak velocity of the COM during the flight phase \((F(3,45) = 31.88, p < .001, \text{Fig. 18})\). For vertical displacement of COM there were significant differences between all conditions with the trend of decreased displacement with increasing frequency. For average velocity of the COM there were significant decreases from MS to MM, MS to MF, MP to MM, MP to MF and MM to MF and for peak velocity there were significant decreases with increasing frequency between all conditions.

![Vertical COM Displacement](image)

Figure 16: Mean maximum COM displacement during stance phase for each metronome condition \((* \text{ denotes significant difference at } p < .05)\).
Figure 17: Mean and standard deviation of the average velocity of COM during stance phase for each metronome condition (* denotes significant difference at p < .05).

Figure 18: Mean peak velocity of the COM during flight phase across all frequency conditions (* denotes significant difference at p < .05).
COM range of motion in the horizontal direction increased as hopping frequency decreased. Statistical analysis found significant differences in the AP range of motion (F(3,45) = 12.83, p < .001, Fig. 15a) and ML range of motion (F(3,45) = 28.60, p < .001, Fig. 15b) during stance phase. Post-hoc analysis revealed significant decreases from MS to MP, MS to MM and MS to MF in the anterior/posterior and MS to MP, MS to MM, MS to MF, MP to MM, and MP to MF in the medial/lateral.

Discussion

Similar to young adults, typically developing preadolescents aged 5-11 years old were able to manipulate their hopping frequency. To increase hopping frequency the subjects increased leg stiffness and to decrease hopping frequency the subjects decreased leg stiffness. At
the higher frequency conditions the subjects did not hop as high, did not displace their COM as much during stance, and traveled less distance between hops.

Compared to young adults, preadolescents had a faster preferred frequency on average at 2.55 Hz. The average preferred frequency of young adults is found somewhere between 2.0-2.2 Hz (Austin et al., 2002; C. T. Farley et al., 1991; Ferris & Farley, 1997; Jones & Watt, 1971). Interestingly, the mean cued frequency for the MS condition of 2.12 Hz fell within the young adult preferred frequency range. The 13 subjects analyzed in this study accurately hopped to the beat of the metronome at the MP and MM conditions with less than 5% deviation on average. They were also able to decrease their hopping frequency for the MS condition and increase their frequency above the MM frequency when completing the MF condition. On average, the subjects were less accurate at matching the metronome at the MS and MF conditions. These findings support our first hypothesis in that the frequencies farther away from the subjects’ preferred frequency were the most difficult.

The elevated preferred frequency of the preadolescents influenced the increased difficulty of the MF condition and, for some, made the MF condition physically impossible. For example, the maximum MF condition for one of the subjects, based on that subject’s preferred frequency, was 4.48 Hz or 268.8 hops per minute. Therefore for this subject, and two other similar subjects, the MF condition was not completed. Another indicator of the difficulty of the MF condition was the average maximum increase across subjects, which was only 28.59%. The target maximum increase for each subject was a 40% increase. Accuracy at this percentage increase has not been studied in young adults. An increase from 2.54 Hz to 3.0 Hz (>18%) was completed by young adults with an average deviation of 2.5% (Moran et al., 2013). This
accuracy would be better compared to the MM condition in which our preadolescent population completed the condition with an average of 3.65% deviation.

However, it was the MS condition that had the greatest average difference of 11.33% from cued to actual. The MS condition may have proved the most difficult because at slower than preferred frequencies hopping no longer followed a mass-spring model (Austin et al., 2002). The mass-spring model allows the subject to conserve and utilize energy from the landing (C. T. Farley et al., 1991). When looking at the leg stiffness charts for each condition the loss of stored elastic energy is apparent in the MS condition (Fig. 19). During stance phase, the center of mass continues to move downwards and ground reaction force plateaus and even decreases. This loss of stored energy requires the subjects to put more energy into the system in order to propel themselves upwards into the next hop.

A) B)
Figure 19: Representative charts of ground reaction force normalized to bodyweight over displacement of center of mass for a singular hop from a A) MS trial, B) MP trial, C) MM trial, and a D) MF trial. All conditions except MS illustrate a mass-spring model because as center of mass displacement increases, force increases (C. T. Farley et al., 1991).

A previous study had young adults change their frequency from a self-selected 2.54 Hz to 2.3 Hz (~10% decrease) and they were able to match the slower frequency with an average deviation of only 2.6% (Moran et al., 2013). This is less of a decrease than what was studied and so no direct comparisons can be made. However, this percentage decrease is within the average maximum decrease shown by the preadolescent subjects of -14.87%. So it can be stated that preadolescents can decrease their frequency, on average, more than 10% and could potentially be just as accurate as the young adult population.

**Leg Stiffness**

Our hypothesis for leg stiffness was supported in that as frequency increased, the subjects increased their leg stiffness. Both methods of calculation, center of mass marker and double integration of force plate data produced similar results and significant differences. Significant
differences were found between all conditions. These results demonstrate that at a preadolescent age the ability to change leg stiffness is not only a prevalent motor strategy, but used to hop at different frequencies. Preadolescents were also able to decrease their leg stiffness in order to hop at a slower than preferred frequency. As stated before, hopping slower than preferred frequency no longer follows a mass-spring model (C. T. Farley et al., 1991). So maintaining leg stiffness would not be an effective strategy to hop at the slower frequency. Since the subjects did hop at the slower frequency they demonstrated their ability to reduce leg stiffness and incorporate other motor strategies to accomplish the task.

It has been shown that the ability to effectively stiffen the leg to hop at different frequencies is found in a young adult population by numerous studies (Austin et al., 2002, 2003; C. T. Farley et al., 1991; Moran et al., 2013). The results of varying leg stiffness across frequencies confirm the presence of this motor ability in a preadolescent population. This motor ability could carry over to many forms of locomotion including walking, running, bounding and jumping. Increased leg stiffness has been shown to increase velocity of movement, increase jumping height, increase running economy, and lessen the risk of soft tissue injury (Butler et al., 2003). The knowledge of the presence of the ability to stiffen the leg in a typically developing preadolescent population will be beneficial when studying children with movement disabilities such as Down syndrome, autism spectrum disorder, or cerebral palsy.

**Toe Marker Movement**

Hopping height displayed a significant trend; as hopping frequency increased hopping height decreased. Post-hoc analysis of hopping height revealed significant differences between all hopping conditions except MM and MF. It seems changing hopping height to match different frequencies is an effective strategy that becomes less useful as frequency continues to increase.
There is certainly a limit to how low a subject can jump and still leave the ground at the higher frequencies. The movement of the COM in the vertical direction during stance phase could provide insight into a more effective strategy at the faster frequencies.

Based on the results from hopping height, the peak vertical velocity of the toe marker during flight phase can be understood. Similar to hopping height there was a significant trend of increased peak velocity as frequency decreased. There were also post-hoc significant differences between the low frequency (MS and MP) and high frequency (MM and MF) metronome conditions. The peak velocity occurred just before landing so gravity was the main accelerator during the downward portion of the flight phase. Peak velocity would then be directly influenced by how high the subject’s toe got off the ground. So peak velocity was expected to have similar results as hopping height.

Our hypothesis for horizontal displacement of the toe marker was not supported as the trend across variables pointed to a decrease in movement as frequency increased. The AP and ML directions illustrated a pattern of an increase in displacement at the lower frequency conditions. Significant decreases of displacement of the toe marker were found between the low frequency and high frequency conditions in both directions. These differences could signify either less or an increased control with a change in frequency. At the high frequency condition the subject may have needed to limit fluctuations in the horizontal in order to vertically rebound quickly into that next hop. In comparison, the low frequency condition may have challenged the subjects’ balance and lead to increased displacements between hops. The difficulty of the MS condition, because it does not follow a mass-spring model, has already been discussed and reinforced with the average deviation from cued frequency across subjects. The difference between conditions could also be explained by the difference in hopping height. The subjects
spent less time in the air during the MM and MF conditions and therefore had less time to travel in the AP or ML direction during the flight phase.

There were no significant differences across frequencies for the range of motion of the toe marker or hopping area. These results point to the maintenance of a similar hopping boundary across frequencies where the subjects corrected their horizontal displacements by hopping in the opposite direction of the preceding hop. However, there potentially could be an insight to different motor strategies considering the displacements between hops decreased with frequency. The subjects completed a greater number of smaller displacement hops at the higher frequencies within the same range and area as fewer but more ground covering hops at the lower frequencies. This relationship seems to illustrate that either the ability or priority to correct horizontal displacements by hopping in the opposite direction decreased with frequency. The range and area results may have been influenced by the instructions to hop on a central location, the force plate. However, it would seem unlikely since the largest mean hopping area for any condition was 565.03cm$^2$ (MS) and the total area of the force plate was 1851.28cm$^2$. The subjects had the possibility of hopping within a larger area than what they maintained for each frequency.

**Center of Mass Movement**

In the vertical direction, the trend of increased movement at the lower frequency conditions was prevalent. The average maximum COM displacement during stance phase significantly decreased across all conditions as frequency increased. This variable signifies how much the subject crouched before going into the next hop. These results suggest that as frequency increased, the subjects held a more erect posture during stance phase of hopping. At the low frequencies the subjects crouched more during stance phase. The increased lowering of
the COM could be an attempt to increase the storage of elastic energy to use for that next hop. However, during the MS condition it has been discussed how energy is lost because the subject no longer acts as a mass-spring model. So at this condition it may be plausible to understand the increased COM displacement as an attempt at balance control. The average COM velocity in the vertical direction further illustrates the small amount of position change at higher frequencies. At the lower frequencies there were greater velocities. Even though the subject spent less time in stance phase at the higher frequencies there was less position change and therefore lower average velocities.

The last variable in the vertical direction was peak velocity during flight phase. One would expect similar results as the peak vertical velocity of the toe marker and the same trend was evident. With an increase in frequency there was a decrease in peak velocity. Contradictory to the toe peak velocity, the peak velocity of the COM produced post-hoc significant differences between all conditions. This could be explained because while in the air the subject could manipulate their COM with their upper body. We attempted to limit the role of the trunk by having the subjects place their hands on their hips. However, these discrepancies illustrate that the subjects were potentially changing the location of their COM while in flight.

Our hypothesis for the center of mass movement was also not supported in that, similar to the toe marker results, there was increased movement and velocities at the lower frequencies. During stance phase there were significant differences of the COM range of motion in the AP and ML planes. The greatest range of motion of the COM was found in the MS condition. The combination of the extended time in stance phase and increased COM range of motion at the MS condition suggests the necessity for more balance control. During the MS condition the subject cannot immediately rebound into that next hop but instead must land, control their sway and then
time their next takeoff. This increased movement of the COM further illustrates the difficulty of the MS condition.

**Limitations and Future Studies**

There were a few limitations to the study. Our study indicates that preadolescents, on average, have the ability to maximally increase their hopping frequency by 28.59% of their preferred frequency. They also, on average, have the ability to maximally decrease their hopping frequency by 14.87%. So we would suggest changing the frequency conditions by +15% (MM), +30% (MF) and -15% (MS) in future studies which would be similar frequency changes as young adults, who have been shown to have lower preferred frequencies. Preadolescents would be better able to complete these conditions, including the MF condition, which was impossible for some subjects.

Another limitation to our study is the use of leg stiffness calculation for the MS condition. At the MS condition the subjects did not follow a mass-spring model. This was consistent with previous findings of frequencies below the preferred frequency (C. T. Farley et al., 1991). Leg stiffness is an accurate calculation of movements that follow the mass-spring linear relationship of force and displacement. Therefore, our leg stiffness calculation for the MS condition overestimates the actual stiffness of the leg (Austin et al., 2002). This overestimation does not impact our results, however, because we still found significant differences between MS and the other conditions for both methods of calculation. Future studies should attempt to analyze more accurate methods to calculate leg stiffness at frequencies less than preferred.

Joint angle kinematics should be studied in the preadolescent population, especially the ankle joint. Torsional stiffness of a joint can be measured by calculating the angular
displacement for a given moment about the joint (Claire T. Farley et al., 1998). Therefore, torsional stiffness of a joint could change and influence leg stiffness. It has been found that the ankle joint is the primary joint influencing leg stiffness for hopping in young adults (Claire T. Farley & Morgenroth, 1999). Comparing the joint kinematics in a preadolescent population would further our understanding of the development of hopping motor control strategies.

Future studies should also incorporate the use of electromyography (EMG) of the lower leg muscles. A previous study looked at muscle activity of the soleus and medial gastrocnemius in neurologically normal men during two legged hopping at different frequencies (Funase et al., 2001). This study attempted to observe the muscle activity in anticipation of landing, muscle activity in response to stretch reflexes, and the muscle activity that contributes to take off. It was found that the medial gastrocnemius would turn on consistently before landing in anticipation but the soleus did not. The soleus muscle would utilize the short-latency reflex response triggered by landing in order to propel the subject into the next hop (Funase et al., 2001). It would be beneficial to understand if preadolescent children have the same muscle activation pattern.

Finally, understanding the capabilities of healthy, typically developing preadolescents and their motor strategies and muscle activation patterns will aid in the understanding of preadolescents with motor disabilities. The observations made in this study can be applied to future studies that look at children with disabilities. The presence or absence of some of the motor strategies and capabilities demonstrated by the typically developing preadolescent population could provide insight when developing beneficial therapy for those children with disabilities.
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References


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Appendix A: Informed Consent

Georgia State University
Department of Kinesiology and Health

Parent Permission Form

Title: The effects of frequency on single-leg hopping in typically developing preadolescents
Student Principal Investigator: Matthew Beers, B.S.
Faculty Advisor: Jianhua Wu, Ph.D.

I. Purpose:

We invite your child to participate in a research study. We will ask your child to hop freely and
to the beat of a ticking device. We aim to study how children hop at different paces. We hope to
use this knowledge to understand the development of hopping.

II. Procedures:

We will recruit 20 children including both boys and girls at 6-11 years old. Your child will come
to the lab once for about 90 minutes. We will provide free on-campus parking. Upon arrival
your child will meet our research staff and explore our lab. We will explain this project to you
and your child in detail. You and your child are free to ask any questions throughout the study.
We will tell your child that he/she may drop out of the study at any time. We will obtain a signed
parent permission form from you and a verbal consent from your child in the lab.

Your child will be wearing a bathing suit or compression shorts. We will use a scale with height
rod to measure your child’s height and weight. We will use a tape measure to record your child’s
limbs’ lengths and widths. We will place 35 shining stickers on your child to observe his/her
hopping movement. Our cameras will track these shining stickers. Your child will hop on one leg
with arms across the chest on a special plate. In the first three trials your child will hop at their
comfortable pace. Your child will then hop to the beat of a ticking device at the same pace and
27% and 45% faster. Your child will hop for 20 seconds at each pace. Your child will have
enough rest between trials. We will collect three trials in each condition. We will remove the
markers from your child at the end.

III. Risks:

Hopping on one foot presents minimal risk for injury. We will give your child enough rest to
prevent fatigue and stress on the foot. A researcher will be nearby in case your child loses his/her
balance.

IV. Benefits:

Your child will not directly benefit from participating in this study. The goal of this study is to
obtain data on the development of hopping. We will use this knowledge for future studies to
understand the development of hopping in special populations with motor control deficiencies.
This shall help us develop appropriate physical intervention for them.

IRB NUMBER: H14287
IRB APPROVAL DATE: 02/04/2014
IRB EXPIRATION DATE: 02/03/2015
V. Voluntary Participation and Withdrawal:

Participation in this research study is voluntary. If your child decides to be in the study and changes his/her mind, he/she has the right to drop out at any time.

VI. Confidentiality:

We will keep your child’s records private to the extent allowed by law. We will use a study identification number (Subject 01) and not your child’s name and initials. We will store your child’s information on a password-protected computer in a locked office. Only the principal investigator and research assistants will have access to them. Results of this study will be reported in group form. Your child will not be identified personally. Information may also be shared with those who make sure the study is done correctly (e.g. GSU Institutional Review Board and the Office for Human Research Protection).

VII. Contact Persons:

Contact Matthew Beerse at 719-238-5569 or mbeersel@student.gsu.edu or Dr. Jianhua Wu at 404-413-8476 or jwu11@gsu.edu if you have questions, concerns, or complaints about this study. You can also call if you think your child has been harmed by the study. Call Susan Vogtner in the GSU Office of Research Integrity at 404-413-3513 or svogtner1@gsu.edu if you want to talk to someone who is not part of the study team. You can talk about questions, concerns, offer input, obtain information, or suggestions about the study. You can also call Susan Vogtner if you have questions or concerns about your child’s rights in this study.

VIII. Copy of Consent Form to Subject:

We will give you a copy of this consent form. If you are willing for your child to volunteer for this research and be video recorded, please sign below.

Please indicate if the principal investigator has permission to contact you for future studies. Y/N

Minor Subject of Research

________________________
Printed Name

Parent or Legal Guardian

________________________  __________________________  __________________________
Printed Name    Signature    Date

Principal Investigator

________________________
Printed Name

________________________
Signature

________________________
Date

GSU APPROVED

IRB NUMBER: H14287
IRB APPROVAL DATE: 02/04/2014
IRB EXPIRATION DATE: 02/03/2015
Appendix B: Child Assent Form

Georgia State University
Department of Kinesiology and Health

Title: The effects of frequency on single-leg hopping in typically developing preadolescents
Student Principal Investigator: Matthew Beerse, B.S.
Faculty Advisor: Jianhua Wu, Ph.D.

Child Assent Script

We will play some fun games with you. First, we will measure your height and weight to see how big and tall you are. Then we will attach some shiny stickers to you. We will use special machines to see how you hop. The first round of the game you will hop on one leg with your arms hugging across your chest. The second round of the game you will hop to the sound of a beeping machine. Every time you hop to the sound of the beeping for 20 seconds, you will receive a sticker. At the end of the game we will count how many stickers you have.