

An Analysis of Infant Bouncing at Different Spring Frequencies

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NOTE:

This thesis is assembled in manuscript format with two manuscripts. The first section consists of a general introduction followed a by review of literature and short methodology. As both manuscripts outline the methods of this study, Chapter 3 only contains additional information that was not included in the manuscripts. Two manuscripts, *Infant Bouncing at Different Spring Frequencies* and *Infant Bouncing: Leg Movements and Mechanism of Control*, are included in the following chapters of the thesis and are intended to be submitted for publication. Finally, a short general discussion outlining the manuscript findings and future recommendations is included.

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ABSTRACT

Infants explore environments through repetitive movements which are constrained or facilitated by the environmental context. The current studies analyzed how typically developing infants bounced in four environments that differed by system natural frequency. Four pre-walking infants (age 9.7 months \pm 1.8) were placed in four spring conditions with natural spring frequencies of 0.9, 1.15, 1.27 and 1.56 Hz. All infants bounced above the natural spring frequency in all conditions suggesting that they do not solely behave like a mass-spring system. Two patterns of bouncing adaptations were identified. Three infants regulated bounce frequency, while one infant regulated the percentage of time on the ground. When infants matched their bounce frequency to the natural frequency, trunk vertical displacement and joint ranges of motion decreased across conditions and demonstrated a shift from non-spring like to circular spring-like phase planes. Moderate to high correlations were found for inter- and intra-limb coordination. Conversely, when an infant regulated time on the ground, trunk vertical displacement and joint ranges of motion remained the same across conditions and inter- and intra-limb correlations were low to moderate. Phase planes remained circular spring-like for this infant. Asymmetrical loading patterns and decreasing vertical ground reaction forces were found in all infants suggesting that a timing component is always regulated. The difference in bouncing pattern may be indicative of different bouncing skill level.

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CHAPTER ONE

INTRODUCTION

During early motor development, one of the first rhythmical movements is supine kicking (Thelen, 1979), as infants begin to kick in a supine position as early as one month of age (Thelen, 1981). Infants gain enough head and neck postural control to sit alone at approximately four to nine months of age. As early as five months, they are able to support their body weight and pull themselves to standing (Haywood & Getchell, 2001, p 97; Piek, 2006, p 40). Throughout this period, infants explore their environments through repetitive movements and manipulate different degrees of freedom to reach their targeted motor sequence. Less understood is what happens when infants are placed in an environment where partial body weight support is provided and their legs are left to explore when they come in contact with the ground.

Dynamical systems theory has been used in various studies to describe infant development. These studies have used different experimental paradigms to demonstrate the motor abilities and coordination of infants in stepping (Thelen, 1986; Thelen, Fisher, & Ridley-Johnson, 1984), kicking (Angulo-Kinzler, Ulrich, & Thelen, 2002; Fetters, Chen, Jonsdottir, & Tronick, 2004; Heriza, 1988a; 1988b; 1991; Jensen, Thelen, Ulrich, Schneider, & Zernicke, 1995; Jensen, Ulrich, Thelen, Schneider, & Zernicke, 1994; Piek & Gasson, 1999; Thelen, 1995), reaching (Clearfield, Dineva, Smith, Diedrich, & Thelen, 2009; Smith & Thelen, 2003) and bouncing (Foo, Goldfield, Kay, & Warren, 2001; Goldfield, Kay, & Warren, 1993; Vallis, 1998).

Dynamical systems theory is based on the principle of self-organization, where movement patterns result from the interactions of various subsystems such as sensory,

perceptual, integrative neural components, cognition and motivation subsystems (Heriza, 1991; Thelen, 1995). Some of the earliest applications of dynamical systems theory were to explain a series of studies on the infant stepping reflex. Stepping rate was first examined in infants at two, four and six weeks of age and the study found that stepping rate decreased with age but also as the infant's leg mass increased. The highest percentage of leg mass increase occurred between two to four weeks of age and stepping rate decreased as a result of additional leg mass. When weights were added to 4-week-old infants' legs, stepping rate subsided; however, stepping rate increased when the legs were submerged in water (Thelen et al., 1984). Infant stepping was indeed not a reflex, but a response elicited by environmental change.

Movement behaviours are influenced by task constraints. Research has demonstrated that task specific actions can be directed by verbal instruction (Arampatzis, Brüggemann, & Klapsing, 2001; Balasubramaniam, Wing, & Daffertshofer, 2004; Dufek & Bates, 1990; Jeka & Kelso, 1995; Kay, Kelso, Saltzman, & Schöner, 1987; Kelso & Jeka, 1992), visual or auditory feedback (Angulo-Kinzler et al., 2002; Prapavessis, McNair, Anderson, & Hohepa, 2003), or other environmental constraints (Jensen et al., 1995; Jensen et al., 1994). Movement amplitudes of bimanual oscillations have been shown to decrease when adult subjects maintain synchrony with an auditory metronome at increasing frequencies (Jeka & Kelso, 1995; Kay et al., 1987). Infants were able to learn motor solutions by reinforcing very specific leg movement patterns of knee flexion and extension with the aid of a mobile that consisted of a computerized biofeedback system and that moved when infants produced the specific leg movements (Angulo-Kinzler et al., 2002). Jumping off a platform from greater heights and to further distances resulted in increased ground reaction forces being generated (Dufek & Bates). However, when individuals were

instructed to decrease the sound made when the foot hit the ground, the ground reaction forces decreased (Prapavessis et al., 2003). Other experiments that contain mechanical components place an environmental limitation on motor behaviours. Placing an infant in a Jolly Jumper and allowing the infant to learn to bounce is an example of an environmental limitation of a bouncing task (Goldfield et al., 1993). The aforementioned research examples highlight various environmental constraints in task-specific movements. The current study investigates how infant bouncing changes as a function of environmental constraints using four different spring frequencies. Kinematic and kinetic data were used to understand how infants adapt to environmental changes and to identify underlying mechanisms that explain their global bouncing behaviour.

There are limited data on infant bouncing. The development of infant bouncing has been previously investigated longitudinally where infants were exposed to a single spring over the experimental period (Goldfield et al., 1993). Skill levels have been characterized based on analysis of a minimal dataset (Vallis, 1998) and limited data from one infant demonstrated that an infant could bounce at different frequencies when the spring parameters were changed (Foo et al., 2001). To our knowledge, no study has investigated infant bouncing across multiple spring conditions using cycle-to-cycle analysis. In order to understand bouncing behaviour and the adaptations of this behaviour, it is essential to obtain continuous bouncing data from infants who already know how to bounce. Furthermore, measuring variables such as bounce frequency, bounce height, vertical ground reaction forces and joint kinematic data is important for understanding the adaptation abilities of typically developing infants and whether and how changes in the spring parameters affect infant bouncing behaviour.

QUESTION

Do infants bounce at the natural frequency of the system when the environmental constraints of the system are altered?

PURPOSE OF THE STUDY

The purpose of the study was to determine whether and how infants would alter bouncing behavior in response to changes in the characteristics of the spring from which they are suspended. Changes in behavior were described using movement kinematics and kinetics. Mechanisms underlying changes in behavior were described using visual analysis of joint angle-joint velocity relationships. A dynamical systems theoretical framework was used to explain the changes in bouncing behaviour.

NULL HYPOTHESES

Primary hypothesis The bounce frequency of each infant will not be significantly different across spring frequencies. This will indicate that infants do not adapt to their environments to behave like the mechanical system and do not bounce at the resonant frequency of the system.

Secondary hypotheses Bounce kinematics: The duration of each bounce when on the ground and the amplitude of each bounce will not be significantly different across spring frequencies.

Bounce kinetics: The peak vertical ground reaction forces of each bounce will not be significantly different across spring frequencies.

Lower extremity kinematics: Ankle, knee and hip joint ranges of motion and inter- and intra-limb coordination will not be significantly different across spring frequencies.

LIMITATIONS

The subject population for this study consisted of four infants aged 6- to 12-months with previous bouncing experience and no known pathologies. Of the four infants, three were male and one was female. Therefore, the results of this study are limited to this population. As an equal number of males and females were not included, sex-based differences could not be analyzed. The study analyzed bouncing at 4 natural spring frequencies for 15 minutes in each condition.

Joint kinematics were determined by external markers attached to anatomical landmarks on the skin, over the clothing and on the bouncing harness of the participants. Directly attaching the markers to bone provides the most accurate measurements of motion of that bone; however, the markers are used to represent bones that are assumed to be rigid bodies. A limitation to this method is that markers move independently of each other and cause error in subsequent kinematic calculations (Hamill & Selbie, 2004).

DELIMITATIONS

Typical development for infants was verified with the parents through simple questionnaires (Appendix H, Infant Data Sheet), thus typical development was subjective and not quantitatively analyzed.

DEFINITION OF TERMS

The following are definitions of concepts and terms that are used throughout this document.

Bouncing A complete bouncing cycle required vertical displacement in which the knees were flexed and extended, where the body moved towards and away from the floor, in conjunction with hip and/or ankle flexion and extension and both feet lifted off the floor during the bounce cycle. Bounce sequences were recognized if bilateral symmetrical movements (both legs moving together) or asymmetrical out-of-phase rhythmical movements (one leg slightly lagging the other) occurred in the lower extremities where feet were on and off the ground together. Bilateral asymmetrical movements, such as stepping, were excluded from bouncing sequences. Skilled bouncing was characterized as symmetrical leg movements, which were well coordinated, and as a result were able to adapt to the changes of the system.

Bouncing bout A bouncing bout was defined as a continuous series of a minimum of six bounces with no pauses between bounces. A pause was defined as the cessation of vertical displacement at the knees where infants no longer moved upwards and downwards, and where the feet did not lose contact with the floor.

Bounce period The duration of a bounce in seconds.

Bounce Frequency The number of bounces per second, i.e., the inverse of bounce period.

Spring System Harness A modified Jolly Jumper harness hung from a bar with two parallel straps. The harness was attached to a support with a link chain.

CHAPTER TWO

REVIEW OF LITERATURE

General Overview

The effects of different spring stiffnesses on bouncing behaviour in typically developing pre-walking infants were examined. The following review of literature encompasses three sections of the different research areas that are significant to the research question of this study. The first section addresses the motor control theory, dynamical systems, which will be used as theoretical framework in the current study. The second section outlines coordination dynamics of the upper and lower extremities from different experiments when environmental parameters are manipulated. Lastly, the third section examines previous infant bouncing research.

Dynamical Systems Theory

In 1984, Thelen, Fisher and Ridley-Johnson challenged previously well-accepted theories of infant development with an elegant series of studies of the stepping reflex. Early explanations of the disappearance of the stepping reflex were heavily based on the argument that maturation of the central nervous system (CNS) inhibited the expression of the reflex (McGraw, as cited in Thelen, 1995, p 81). Thelen et al. (1984) were the first to propose that the disappearance of the stepping reflex was due to rapid leg mass gain during four to six weeks of age, instead of the development of inhibitory nervous system tracts as otherwise argued by Forssberg and McGraw (as cited in Gershoffk-Stowe & Thelen, 2004, p 107). With this work, Thelen and colleagues expanded the study of infant development

from reflex and maturational theories and applied dynamical systems theory to motor development.

Dynamical systems theory (DST) originated from Bernstein's system theory (Heriza, 1991; Shumway-Cook & Woollacott, 2001, p 17). Bernstein looked at the whole body as a mechanical system and suggested that it was impossible to understand the neural control of movements without understanding the external and internal forces acting on the body. Initially, DST was based on a fundamental principle of self-organization. This principle states that there is no need for a higher center sending out instructions or commands to achieve coordinated movements, as described in maturational and hierarchal theories (Heriza, 1991; Shumway-Cook & Woollacott, 2001, p 18). Instead, movements occur as a result of individual parts interacting without the need for specific motor programs within the nervous system. Initially, DSTheorists would have indicated that neural contributions were trivial when it came to motor development; however, this is not the current stance of dynamical system theorists (Heriza, 1991; Thelen, 1995; Shumway-Cook & Woollacott, 2001, p 20).

The current interpretation of DST integrates different subsystems, which include environmental, sociological, psychological, perceptual processes, cognition, motivation, physiological and neural. Neural subsystems are responsible for generating muscular patterns that lead to joint synchrony, postural control such as balance and equilibrium, muscle strength, and extensor/flexor control. These subsystems assemble to produce human movements (Heriza, 1991; Thelen, 1995). According to Heriza (1991), DST acknowledges that the central nervous system is important to motor behaviour; however, it is not the sole factor. Therefore, a movement pattern results from the interaction of various subsystems,

which assemble from the demands of the task and the environmental circumstance of the system.

Thelen et al. (1984) argued that the stepping reflex does not disappear as a function of maturing inhibitory tracts in the motor cortex, as the reflex reappeared in older adults and in stroke patients. Rather, the reflex could be elicited through appropriate environmental constraints. For example, Thelen et al. used a longitudinal design to look at the stepping rate of infants at two, four and six weeks of age. Their findings demonstrated that stepping rate decreased as infants aged, especially between two to four weeks, which corresponded to the period of highest percentage of fat mass increase. To further validate their theory, they manipulated infant leg mass and hypothesized that stepping rate would change if infants were submerged in water or if weights were added to their legs. The experiment demonstrated that stepping rate increased when 4-week-old infants were submerged in water, simulating mass reduction, and decreased when weights were added (Thelen, Fisher, & Ridley-Johnson, 1984). The changes in stepping frequency occurred without CNS maturation supporting a dynamical systems explanation and challenging a maturational standpoint. The disappearance of the stepping reflex resulted from the rate-limiting factor of leg growth.

When applied to development, DST meets two assumptions, multicausality and nested timescales (Smith & Thelen, 2003). Multicausality describes developing organisms “as complex systems composed of many individual elements embedded within, and open to, a complex environment” (Smith & Thelen, 2003, p343). Instead of having a single element, such as a program or a hierarchical agent that will produce an organized pattern, DST indicates that self-organization occurs from independently driven components, as a solution to a problem in the environment. Consequently, movement patterns arrive at a

solution where variability of a particular movement is low and forms more efficient movement patterns. The concept of self-organization and efficient solutions are referred to as *soft-assembly* and *attractor states* in the DST, respectively (Thelen, 1995). An attractor state is a movement pattern that is preferred by the system when starting at various initial conditions, or when a system is perturbed (Heriza, 1991; Thelen, 1995). The longer a specific behaviour is performed, the deeper the attractor state.

To reach an attractor state, however, all components of a system must reach optimal functioning; thus, a rate-limiting factor cannot be present (Thelen, 1995). A rate-limiting factor prevents or limits preferred self-organization. Rate-limiting factors are time dependent, as development progresses, the rate-limiting factor changes. For example, the stepping reflex (response) disappears when there is insufficient leg muscle strength to lift the additional leg mass gained in the first few months (Thelen et al., 1984). Several months later, although typically developing 7-month-old infants contain the postural support to control their trunk, they are unable to walk independently until adequate leg strength develops. In these two examples the rate-limiter is leg muscle strength; however, they apply to different developmental situations. This leads to the second assumption, nested timescales, which specifies that behavioural changes occur over different periods of time (Smith & Thelen, 2003). Using the same two examples above, to overcome the rate-limiter, leg muscles require different durations to lift the additional mass and elicit the stepping response in comparison to the time period the infant needed to acquire adequate leg muscle strength to begin walking.

According to Heriza (1991), DST suggests that movement patterns are favored and not mandatory, and occupy preferred regions of their state space. The term “preferred” implies that there is no prescription in advance for a movement pattern; instead, the system

desires to move in a specific way (Heriza, 1991). The assumption negates that movement patterns are hard-wired or programmed, suggesting that movements are flexible and softly assembled. With regards to infant kicking, Heriza suggested that preferred movement patterns could be graphically represented with joint angle by joint angle plots (angle-angle plots) of the same and different limbs and joint velocity against the joint angle (phase plane plots) of the same limb. The former graph gives insight to inter- and intra-segmental coordination, while the latter demonstrates a picture of a control mechanism (Winstein & Garfinkle, 1989). If movement patterns are stable, i.e., in an attractor state, then the movement pattern occupies a preferred space in the plot (Heriza). The aforementioned components can be used to analyze other types of infant movement such as infant bouncing.

Dynamical system theory can be used to explain infant bouncing, as it is a developmental behaviour that occurs as a result of infants discovering the given task. If infants learn the low-dimensional dynamics of the bouncing task, it will provide further evidence to support the theory. Infant bouncing studies (Goldfield, Kay, & Warren, 1993) have used DST to explain the infants' behaviour, and have defined low-dimensional dynamics as preferred attractor states where the degrees of freedom of the system are reduced from an infinite number. The infants' inter- and intra-limb joint coordination and bouncing skill will depend on how stable their attractor state is at a given spring frequency. As spring frequencies change, one will be able to identify whether different movement patterns occur and how often they are used.

Motor Behaviours under Environmental Constraints

Oscillating behaviours. In dynamical systems, movement behaviours are a result of multiple components self-organizing to produce a specific behaviour. Studies have shown phase transitions, which represent the transition from one attractor state to another attractor state, when subjects are instructed to perform a task or when they are placed in different environmental conditions. Two studies (Jeka & Kelso, 1995; Kelso & Jeka, 1992) have outlined the coordination dynamics of different limb combinations and the frequencies at which inter-limb movements become unstable resulting in a transition from one attractor state to another. In other studies, the effects of task-specific timing perturbations on various oscillatory movement behaviours have been demonstrated (Balasubramaniam, Wing, & Daffertshofer, 2004; Jeka & Kelso, 1995; Kay, Kelso, Saltzman, & Schöner, 1987).

Coordination dynamics of a multiple component system using arms and legs were analyzed through various experiments (Kelso & Jeka, 1992). Participants sat in a multiarticulator coordination apparatus, a device that attached to the ankle and wrists, allowed full flexion and extension at the knee and elbow and permitted simultaneous measurement of position and velocity of the four limbs. The researchers investigated whether all combinations of flexion and extension movements among the arms and legs were equally stable. For each subject, a total of 12 limb conditions were tested. The combinations included homologous (e.g., right and left arms) and ipsilateral limb pairs (e.g., right arm and right leg). Two starting positions (modes) of coordination, 0° and 180° out of phase were used. Coordination patterns were characterized in terms of direction of motion as “same” or “different”. “Same” referred to movement of the segments in the same direction despite the muscle groups involved. For example, in homologous pairs (right arm and left arm), “same” corresponded to flexion-flexion and extension-extension about the

elbow. With ipsilateral pairs, such as right arm and right leg, “same” corresponded to flexion of the right arm with extension of the right leg about the knee. Ipsilateral conditions were the least stable and had the greatest relative phase variability, particularly when the limbs were moving in different directions such as when flexion was accompanied by flexion, and extension with extension. Therefore, Kelso and Jeka (1992) reported that difficulty with coordination increased when movements were made with two arms and two legs to one arm and a contralateral leg and then again to an ipsilateral arm and leg.

In the initial study above (Kelso & Jeka, 1992), subjects moved at their own preferred frequency. The experiment was then extended and participants were asked to repeat the same 12 conditions but with the frequency driven using an auditory metronome. Two additional conditions were added that incorporated movement of all four limbs. Movement frequency increased from 1.25 Hz to 3.00 Hz in intervals of 0.25 Hz with 10 cycles per frequency. Transitions in movement coordination were observed in all conditions. When limbs started in “same” mode, subjects switched to “different” mode in 2% of all trials. When subjects began in “different” mode, switching was recorded in 75% of the trials. Kelso and Jeka (1992) demonstrated that the “different” mode conditions were not only more variable than “same” mode, but that stability decreased with increasing cycling frequency. In addition, “different” mode was more variable than “same” mode before a transition. A key finding of this study was the transition behaviour of the limbs. Kelso and Jeka reported that ipsilateral and nonhomologous contralimb pairs switched from “different” to “same” mode conditions at an average frequency of 1.55 Hz and 1.92 Hz, respectively. This implies that at frequencies above 1.5 Hz ipsilateral limbs can only move in the same direction, while the nonhomologous contralateral limbs are stable in either mode until 1.90 Hz.

The phase transitions, which were highlighted in former study (Kelso & Jeka, 1992), demonstrated the digression of one stable movement pattern to another stable movement pattern. An important characteristic of a transition was the increase in relative phase variability. A shift of a stable movement pattern to a less stable or unstable movement pattern where stability is reduced can be measured as increased variability (Piek & Gasson, 1999). Thus, the transition from “same” to “different” was an example of two different attractor states, the former condition more stable than the latter.

Subsequent studies by the same group investigated the effects of increasing inertial masses to an individual limb on the symmetry of coordination dynamics between an arm and leg (Jeka & Kelso, 1995). Participants were placed in the same multiarticulator coordination apparatus; however, only the dominant side was tested. Ten seconds of movement at six frequencies from 1 Hz to 2.25 Hz with 0.25 Hz increments were recorded. All participants began in “different” mode and were instructed to complete a full cycle of movement with each beat of the metronome. Subjects completed three conditions (no load, arm load and leg load), where an adjustable weight of 1.8 kg was strapped to the wrist or ankle. Relative phase, peak-to-peak cycle duration and movement amplitude of each limb within each loading condition were analyzed.

The study found that inertial limb loading did not affect individual limb cycle duration. Mean cycle movement amplitudes for the arm and leg, however, decreased with increasing frequencies with amplitudes under loaded conditions smaller in comparison to the no load condition (Jeka & Kelso, 1995). Both limbs compensated for an additional load to a single limb, as both mean arm and leg amplitudes decreased when a load was added only to the leg. In addition, the study found phase transitions from “different” to “same” in all loading conditions with the variability of the relative phase increasing as the transition

frequency approached, suggesting that pattern changes were a result of loss of stability and not the result of loading condition. The findings in this study (Jeka & Kelso) demonstrate that regardless of environmental changes, such as increasing the weight of a limb, bimanual movements can maintain the required movement frequencies; however, to do so, phase transitions (movement modes) occur and limb movement amplitudes decrease suggesting a change of attractor states. Although this study emphasizes the behaviour in bimanual limb coordination, the transitions between driving frequencies were continuous and not discrete as proposed in the current project.

Similar behaviours were also seen in discrete movement transitions. Kay, Kelso, Saltzman and Schöner (1987) investigated the relationship between space and time using single and bimanual rhythmical movements of one or both hands. Wrist movement amplitudes and peak velocities were investigated as movement frequencies increased from 1 to 6 Hz at 1 Hz intervals. Participants also performed wrist movements at a self-paced frequency in both the single and bimanual condition as well as two bimanual conditions, mirrored—a symmetric mode, and parallel—an asymmetric mode. Self-paced oscillations were equivalent between single hand conditions (1.90 Hz) and mirrored conditions (2.04 Hz), and both of these motions oscillated at higher frequencies than asymmetric oscillations (parallel bimanual condition) at 1.56 Hz. Amplitude differences were not found in the self-paced conditions although wrist movements oscillated at different frequencies. In metronome-paced conditions, subjects were able to match the auditory metronome in single hand and bimanual conditions, and as wrist frequency increased, peak velocity increased and wrist amplitude decreased. In the bimanual conditions, the mirrored condition was more stable than the parallel condition with variability dramatically increasing between 2 and 3 Hz in the parallel condition (Kay et al., 1987). The increase in

variability suggests decreased stability and a transition to an unstable attractor state.

Similar to the work from Jeka and Kelso (1995), as the frequency increased, movement amplitude decreased when movement velocity increased. These changes were evident in phase plane plots, as higher frequency movements appeared to be more sinusoidal, producing more elliptical shaped phase planes (Kay et al.,). Thus, behavioral changes of movement amplitude as a function of velocity can be detected through visual inspection (phase plots).

Both Kay et al. (1987) and Jeka and Kelso (1992) were interested in identifying how oscillatory movements would change as a function of frequency modulation. The studies identified that cycle amplitude decreased as a result of maintaining the required frequency regardless if the change in frequency was continuous or discrete. These studies do not, however, address how different types of oscillatory movement trajectories assist in maintaining timing accuracy.

Balasubramaniam, Wing and Daffershofer (2004) expanded the definition of coordination and timing relative to an external metronome to include motor goals that required non-contact flexion or extension of the finger. Skilled musicians were asked to produce rhythms with their index fingers at different frequencies (1, 1.33 and 2 Hz). The oscillation trajectories of the finger were recorded as subjects oscillated at their own pace, synchronized their finger with peak flexion (fON) or peak extension (eON) occurring on the metronome beat, or syncopated their finger with peak flexion occurring off the metronome beat (fOFF). Asymmetrical movement trajectories and their relations to task and motor goals were analyzed to determine whether there were timing implications.

Non-contact, unpaced finger oscillations produced symmetrical sinusoidal movement trajectories (Balasubramaniam et al., 2004). This indicated that approximately

equal time was spent during flexion and extension. Moreover, although movement asymmetries were found in paced task-specific conditions, all movement trajectories were more symmetrical and smoother as movement frequency increased.

In comparison to the unpaced conditions, flexion was shorter in duration in the flexion task conditions (fON and fOFF), while the extension task condition (eON) produced shorter extension periods. The similarity between the two flexion conditions (fON and fOFF) was evident in the phase plots, whereas eON displayed a 180° shift (non-superimposable mirror image) from the other paced conditions (Balasubramaniam et al., 2004). The researchers proposed that the deviation from symmetry and smoothness was related to the reduction in timing errors. Their results confirmed this, as the greater asymmetry found, the closer it was to synchronous timing in such way that when the finger arrived early and produced large asynchrony, the finger would compensate by a longer return phase that would correct for the asynchrony. They also mention that the reverse would occur, “a late arrival was compensated by a shorter return phase” (Balasubramaniam et al., 2004, p 132). In light of these findings, flexing on the beat and extending on the beat are not equivalent though they contain equal task goals. Thus, with an auditory metronome flexion off the beat syncopation is equivalent to flexion on synchronization albeit at 180° phase shift. Overall, the study demonstrated that asymmetry is strongly associated to timing, as greater trajectory asymmetry led to better timing accuracy.

The findings from the above studies (Balasubramaniam et al., 2004; Jeka & Kelso, 1995; Kay et al., 1987; Kelso & Jeka, 1992) show that task-specific movements generate similar decreasing amplitudes to maintain the timing component of the task. Also, as demonstrated by Kelso and Jeka (1992), to maintain timing in bimanual movements, a phase transition usually occurs from one attractor state to another and is illustrated through

a change in the relationship of segment kinematics. Phase transitions were also detected with increased variability in relative phase. An increase in variability suggests that a less stable or unstable attractor state exists. As seen in all the aforementioned studies, environmental constraints, such as maintaining timing with an auditory metronome, generate specific movement behaviours. To maintain timing may require changes of attractors, which can be detected in kinematic data.

Leg behaviour. In the previous section, phase transitions of oscillating movements from different attractor states were presented. Phase transitions, however, are not the only method of identifying stable versus less stable movement patterns. Another method that allows detection of stable movement patterns includes identifying common behaviours in kinematic and kinetic variables. It has been shown that movement patterns have been driven to attractor states or stable movements by task constraints placed in a system. More specifically, the following section will demonstrate how specific movement patterns were elicited through verbal instruction and environmental constraints of the task.

A few leg movement (Angulo-Kinzler, Ulrich, & Thelen, 2002; Jensen, Thelen, Ulrich, Schneider, & Zernicke, 1995; Jensen, Ulrich, Thelen, Schneider, & Zernicke, 1994) and jumping studies (Arampatzis, Brüggemann, & Klapsing, 2001; Dufek & Bates, 1990; Prapavessis, McNair, Anderson, & Hohepa, 2003) underscore how different motor solutions are modulated to meet task constraints. Examples of modulated motor solutions include, but are not limited to, infants kicking at specific angular thresholds to move a mobile, jumping to produce specific vertical ground reaction forces and jumping to produce specific leg stiffness'. Also, the results of changing environmental conditions on leg kicking and jumping have been investigated. The environmental conditions include changes in body position and distance from the ground, respectively. The following studies

will demonstrate how variables such as joint range of motion and vertical ground reaction force (VGRF) were regulated to meet task constraints.

Infant kicking and jumping movement patterns require similar flexion-extension movements of the lower extremity to those seen in infant bouncing. Bilateral, symmetric movements, which are seen in infant kicking and jumping, have shown more efficient intra-limb coordination when little or no phase relations between movements are present over out-of-phase movements (Kamm, Thelen, & Jensen, 1990). Infant kicking is one of the first forms of leg movement, which is independent of postural development, while landing from a jump requires complete upper and lower body stability. In infants who generate bouncing, the behaviour emerges at some time point between kicking and jumping, and is present once trunk posture develops but prior to independent standing. The spontaneous movements of infant kicking in supine and vertical positions have demonstrated similar joint movements and timing of adult walking (Haywood & Getchell, 2001, p 87). During early months of development, infants begin to coordinate their muscles for future locomotor skills. The development of infant bouncing requires the assembly of head control and trunk weight support, with the addition of symmetrical leg kicking or pushing when their feet come in contact with the ground.

Greater spontaneous kick vigor and synchronicity was recorded from 3-month-old infants who were placed in a vertical posture when compared to supine or angled postures (Jensen et al., 1994). Kicks were not isolated movements but instead a series of movements, and it was theorized that muscle extensor and flexor forces would change with joint angle similar to how spring forces change with the stretching of a spring (Jensen et al.). The study suggested that infants produce spring-like qualities in their legs to initiate

spontaneous kicking and that vertical positioning of the body forced distinct interjoint coordination as a result of gravitational forces applied on the body.

A companion study on spontaneous kicking looked at the age-related differences in limb control between 2 week, 3 month, 7 month-old infants and adult vertical kicking (Jensen et al., 1995). At 7 months of age, infants demonstrated greater knee torque in their kicking patterns, which played a role in their power generation, and a decrease in coupled joint action. Through practice, joint independence and joint specific power generation increased, while greater control of forces produced by the joints was demonstrated (Jensen et al.). These studies demonstrated the motor development acquisitions of the lower extremities. The leg movements of infant kicking are spontaneous, as the infants' kicking was related to the level of arousal rather than a goal directed behavior (Jensen et al.).

Spontaneous infant kicking, an open-loop task, can become a closed-loop task-specific movement under specific conditions. Angulo-Kinzler, Ulrich and Thelen (2002) were interested in determining whether young infants could learn a specific leg motor pattern under task constraints. They attached biaxial goniometers to the knees of 3-month-old infants. The goniometers were connected to a motorized feedback system that moved a mobile hanging above the infant. The mobile moved when infants generated an efficient and exact kicking pattern based on the angular threshold of knee flexion (85°) or extension (35°). Two groups of infants, a flexion and extension group were tested. Angulo-Kinzler et al. (2002) set the relative learning frequency at 1.5, which was established from previous studies (Fagen et al.; Ohr & Fagen; Rovee-Collier et al., as cited in Angulo-Kinzler et al., 2002, p 61). Once the infant was able to generate the appropriate angular range 50% more than at baseline and for at least 2 out of 3 minutes, the task was deemed learned. The learning rate for each 2-minute period was calculated as a ratio of the frequency of

reinforcements over the average baseline frequency of threshold crossing, which is when the mobile did not move. With this setup, researchers were able identify that infants could modify their leg movements with contingent reinforcements.

The results demonstrated that 61% of the infants in the flexion group and 62% of the infants in the extension group learned their respective tasks. Infants in the flexion group took 4.75 minutes while infants in the extension group took 5.30 minutes. Infants in the flexion group learned to move the mobile without significantly increasing the relative leg kick frequency, whereas infants in the extension group did the opposite (Angulo-Kinzler et al., 2002). Regardless of group, infants were identified using two different strategies to move the mobile. Some infants moved with large ranges of knee motion that surpassed the degree threshold while other infants produced small fluctuations of knee joint position near the degree threshold that triggered the mobile continuously (Angulo-Kinzler et al., 2002). Completing the task of moving the mobile under specific parameters was not limited to one method as infants' demonstrated alternate methods that reached the same outcome. However, any single infant used one pattern of knee movement (either large range or small fluctuation) only.

Like kicking, jumping requires the leg to flex and extend; however, jumping also requires propulsion of the body from a surface with either one or both feet and the ability to land and balance when coming in contact with the ground. When landing from a jump, modifying the task goals can alter movement patterns. Prapavessis, McNair, Anderson and Hohepa (2003) examined the effects of verbal instruction associated with the control of joint kinematics and the resulting auditory feedback on decreasing peak VGRF when landing from a jump in children. Prior to each jump, the researcher instructed children with specific foot and leg landing strategies. The children were told to listen to the landing

sound and to use this as feedback about how softly they landed. This study followed a randomized control-experimental design, where the experimental group was given additional instructions, after their baseline testing for three sessions, on proper feet placement and to listen to the sound of their landing. The study also analyzed if the children were able to retain the information by testing them three months later. Following instruction, nine-year-old children were able to significantly reduce their peak VGRFs in comparison to the control group; however, they did not retain the information three months later (Prapavessis et al., 2003).

Although infants cannot be instructed to land more lightly when bouncing to decrease their VGRFs, previous studies have demonstrated that over time infants could learn to increase bounce height (Goldfield et al., 1993) and data from a single infant showed that bounce frequency could also be modulated (Foo, Goldfield, Kay, & Warren, 2001). This suggests that infants can adjust their movements to achieve an optimal bounce height or bounce frequency within their environmental constraints. Although it remains unclear how the task is achieved, one could hypothesize that the forces infants produce as their feet come in contact with the floor and the duration the ground reaction forces (GRFs) will be also modified to regulate the low-dimensional dynamics of the bouncing task.

Another factor that may contribute to the application and the duration of the GRFs is leg stiffness. Similar movements to infant bouncing in adults are drop jumps on a springboard. Arampatzis, Brüggemann and Klapsing (2001), suggested that controlling leg stiffness through the ground contact time can influence the mechanical work and power produced. Similar to Prapavessis et al. (2003), the study examined the effect of verbal instructions on the ability to control the stiffness of the lower extremities and the effect of leg stiffness on the mechanical energetic processes during drop jumps from two heights (20

and 40 cm) onto a sprung surface. In this study (Arampatzis et al., 2001), linear regression equations were used to calculate stiffness. Leg stiffness was approximated using a linear spring, as the relationship between the VGRF and the vertical positional changes of the centre of mass (COM) of the subject with respect to the surface. Ankle and knee joint stiffness were approximated by rotational springs, as the relationship between the joint moment and change in joint angle (Arampatzis et al.).

The study found that leg, ankle, and knee stiffness' influenced the duration of the ground contact, where shorter ground contact times corresponded to higher leg and joint stiffness measures (Arampatzis et al., 2001). Leg stiffness was well approximated using linear springs during the negative phase of the drop jumps as high linearity (r) of .92 to .97 was found between the VGRF and the vertical positional change of the centre of mass at both jumping heights of 20 and 40 cm. Similar to the findings in Dufek and Bates (1990), as leg stiffness increased, VGRFs increased, while the subject's vertical positional change in COM decreased. More importantly, as ground contact time decreased, the amplitude of the change in knee angle also decreased (Arampatzis et al.). A similar jumping study (Dufek & Bates, 1990) investigated the effects of different heights, distances and landing techniques on impact forces. The findings suggested that to decrease VGRFs, the knees are required to be slightly or fully flexed. Therefore, variables such as leg and joint stiffness, the forces produced and the duration of those forces have all been modified to maintain the task goal.

The preceding studies (Angulo-Kinzler et al., 2002; Arampatzis et al. 2001; Dufek & Bates, 1990; Jensen et al., 1995; Jensen et al., 1994; Prapavessis et al., 2003) highlighted that attractor states of specific behaviours can be reinforced or become more stable by manipulating the environmental condition, as seen in infant kicking when postural position

was changed and visual feedback was provided. In jumping, to reach lower VGRFs, joint stiffness' were required to decrease (Arampatzis et al.) by reducing knee joint position to slightly flexed or fully flexed (Dufek & Bates) or reducing the sound of foot-ground contact (Prapavessis et al.). Therefore, environmental changes elicit different attractor states as seen with repeated behaviours in the kicking and jumping studies. Attractor states can also be strengthened producing the same task repeatedly within a constrained environment. These studies have identified important variables, which have yet been explored in the context of infant bouncing and may further explain an infant's capability to learn to a new task such as bouncing in a Jolly Jumper at varying frequencies.

Infant bouncing

When 6-month-old infants are placed in Jolly Jumper bouncers, they may generate bouncing behavior as a consequence of their feet kicking or pushing against the floor. Infant bouncing becomes a closed-loop task-specific movement through the assembly and tuning of an action system (Goldfield et al., 1993). Assembly is the "process of self-organization that establishes a temporary relationship among the components of the musculoskeletal system, transforming it into a task-specific action system such as a kicker" (Goldfield et al., 1993, p 1129). When infants begin to explore a new environment for the first time, their movements are variable, as they have not narrowed down the degrees of freedom to achieve a given task. Sporadic, irregular kicking without continuous bouncing represents an assembly phase in infant bouncing (Berthouze & Goldfield, 2008). Tuning results from movement patterns modifying to specific conditions (Berthouze & Goldfield, 2008; Goldfield et al., 1993). Through tuning, infants will demonstrate less variability in the movement patterns and will refine the degrees of freedom to achieve a task. In the case

of bouncing, the infant would be periodically kicking with variable leg stiffness and flexion-extension cycles (Berthouze & Goldfield). This suggests that when infants are placed in a bouncer, movement patterns are generated by the reaction of their feet against the floor and are refined (or tuned) through practice of the same movements.

The first reported infant bouncing study (Goldfield et al., 1993) intended to outline “an approach to motor learning and development from a dynamical systems perspective” (p 1129) using preliminary infant bouncing results. Goldfield et al. (1993) modeled bouncing as a forced linear mass-spring system. The researchers determined the resonant period, the spring stiffness and damping ratio of the bouncing system and compared the infant’s bouncing oscillation period to the resonant period of the spring. Goldfield et al. suggested that if infants’ legs behaved in the same manner as springs, it indicated that infants could match the resonant frequency of the system and that they had learned the low-dimensional dynamics of the task.

To answer this question, Goldfield and colleagues (1993) recorded the bouncing behavior of eight infants weekly for a minimum period of six weeks. Over the course of their bouncing sessions, all infants demonstrated sequential periods of assembly, tuning, followed by a period of “resonance”, which Goldfield and colleagues equated with peak bouncing. The greatest number of bounces an infant produced per bouncing bout was identified as the infant’s peak of bouncing. The author’s hypothesized that at peak bouncing the infants were bouncing at or close to the resonant frequency of the spring system. Each infant’s peak bouncing frequency was compared to the predicted frequency of a single and a double (spring in series) spring model. The predicted oscillation frequency of the single spring model was 1.25 Hz. This model did not account for the incorporation of leg stiffness and the predicted oscillation frequency was less than the mean bounce

frequency of all infants, which was 1.67 Hz. Conversely, when the peak bouncing frequency of each infant was compared to the predicted oscillation frequency of the double spring model, suggesting that infant's legs behave like a spring that matches the impedance of the external spring, results were visually very close in their figures (Goldfield et al. 1993). As a result, Goldfield et al. suggests that infant bouncing behavior can be modeled or interpreted as two springs in series.

This first infant bouncing study was limited with analysis restricted to the number of bounces, bounce amplitude and bounce frequency within predetermined trial lengths. Given the basic kinematic output, the study, however, provided insightful predictions regarding infant development. The researchers predicted that if infants had learned the low-dimensional dynamics of the task instead of fixed leg stiffness or forcing frequency (the amount of force and timing of the force), the infant should quickly adapt to changes in the task conditions. This implies that any modifications to the system's mass or spring stiffness, which consequently changes the systems' resonant frequency, would require the infant to change their forcing frequency and leg stiffness (Goldfield et al. 1993). Alternatively, if infants could not learn the low-dimensional dynamics of the task but rather a fixed frequency and leg stiffness, Goldfield et al. (1993) suggested the infants would require a long period for adaptation if the task constraints were changed.

A subsequent study was designed to investigate specifically whether infants' learned specific parameter settings or whether they learned the low-dimensional dynamics of the task (Foo et al., 2001). The published results are limited to data from one infant. When the spring constant was changed or additional mass was added to the 8.17 kg infant, the infant demonstrated that, with experience, he could bounce at different frequencies. More specifically, the infant bounced at an initial frequency of 1.63 Hz in the baseline single

spring condition similar to the bounce frequency recorded in the Goldfield et al. (1993) study.

Moreover, the infant bounced at frequencies closer to the predicted (calculated) double spring model than the single spring model (Foo et al., 2001) corresponding to Goldfield et al. (1993) study. The predicted single and double spring model from Foo et al. (2001) coincides with previous values of Goldfield et al. initial condition. The single spring model, which disregarded the infant's leg stiffness, oscillated at 1.26 Hz, while the predicted double spring model that accounted for the infant's leg stiffness was calculated at 1.72 Hz (Foo et al.). As mentioned, the infant bounced at 1.63 Hz in the initial condition. The researchers predicted that a similar proportional reduction in bouncing frequency should occur when an additional mass was added (Foo et al.). Their prediction held true for the predicted single and double spring models with frequencies of 1.12 Hz and 1.49 Hz, respectively; however, when a 2.30 kg mass was added to the system, which would normally result in a reduction of the resonant frequency of the system, the infant bounced at 1.71 Hz. When an additional spring was added, the predicted frequencies were 1.72 Hz and 2.10 Hz for the single and double spring model, respectively. The observed mean bouncing frequency of the infant increased, as expected, to 1.84 Hz when an additional spring was added in parallel to the system (Foo et al.). It was evident that infants always bounced at higher frequencies than the predicted spring model, but lower than the predicted double spring model. An exception was seen when additional weight was added to the infant, as bounce frequency was higher than the predicted single and double spring models. Table 1 summarizes the finding of the two bouncing studies.

Table 1 - Summarized results of bounce frequency and the predicted single and double spring models from two bouncing studies.

	Goldfield et al. (1993)	Foo et al. (2001)		
		Initial condition	+ 2.30 kg	Two springs in parallel
Infant Bounce frequency (Hz)	1.67	1.63	1.71	1.84
Predicted single spring model (Hz)	1.25	1.26	1.12	1.72
Predicted double spring model (Hz)	1.72	1.72	1.49	2.10

Although Foo et al. (2001) demonstrate that the double spring model is similar to the bouncing frequency the infant produces, the researchers do not specify how they calculated the single and double spring model. Additionally, the abstract does not indicate whether the additional spring was added with or without the additional 2.30 kg. Also, it remains uncertain why the infant bounced at a higher spring frequency when a mass was added, as their bounce frequency should have decreased if the infant was intending to match the resonant frequency of the system as Goldfield et al. (1993) hypothesized.

In a recent study, Berthouze and Goldfield (2008) compared commonly used developmental biology principles to dynamical system concepts: assembly, tuning, stable attractor state to optimize robotic modeling of infants. They argue that robotic models of infant bouncing behave similarly to the infants themselves. In robots, assembly was defined as the “self-organization of the collection of oscillators—the adaptation of the connection weights between each oscillator” (Berthouze & Goldfield, 2008, p 33) while tuning resulted from time-constant adjustments of each oscillator. The intent behind the

experiments was to determine whether a robot was capable of modifying its behaviour or whether the robot would organize its behaviour similarly to the longitudinal pattern in infants. The bipedal robot received sensory feedback from four-force sensing resistor sensors placed underneath the feet. The feet and joints of the robot were designed to be passively compliant (Berthouze & Goldfield, 2008). Similar findings to Foo et al. (2001) were observed in the robotic model when maintaining control parameters (mass, stiffness and damping of the system) constant and changing lift-off weight (Berthouze & Goldfield). The sensory feedback the robot receives, the lift-off weight, was improved with low pass filtering. Through phase locking, a process that occurs when two oscillating systems are driven to oscillate at the same period, the robotic model was able to adapt to manual perturbations smoothly. The differences in stabilization between robots and infants can account for the variation seen in individual bouncing frequencies and the predicted single and double stiffness models. Robotic models reached a stable attractor state with alternate phases of freezing and freeing mechanical degrees of freedom of distal freely rotating joints. Infants also demonstrate stabilization, where sustained bouncing occurred at the resonant period of the system; however, the process of reaching a stable attractor state was very different from infant to infant, while robots reached stabilization in the same manner (Berthouze & Goldfield).

An experimental study examined the kinematics and more specifically the phasic relationships between joints in infants with different bouncing skill levels. Vallis (1998) characterized infants as “skilled” bouncers when they were able to produce several bounces without stopping and “less skilled” bouncers when infants demonstrated disorganized movement patterns with sudden stops and starts. “Moderately skilled” bouncers were infants who demonstrated both characteristics. The data showed that skilled bouncers

flexed and extended at the ankle, knee and hip simultaneously while less skilled bouncers demonstrated out-of-phase joint relationships. In agreement with Goldfield et al. (1993), Vallis (1998) suggested that skilled bouncers produce bouncing frequencies at multiples of the spring resonant frequency, while less skilled infants varied and generated a fraction of the resonant frequency and a multiple.

The findings from Vallis' (1998) study showed that the bouncing frequency of less skilled bouncers deviated from the experimentally calculated frequency of the mass-spring system. In contrast to Goldfield et al. (1993), Vallis (1998) reported that skilled and moderately skilled infants produce two types of bouncing behaviour where infants bounce at two distinct frequencies with two different loading patterns using different mechanism of control. Her results demonstrated that infants produced three types of lower extremity phase plane patterns: a smooth rounded pattern (approximating the phase plane of a spring), a cusp pattern (a limaçon that approximates the phase plane of a spring with a forcing function), and a combination of both. Infants who bounced at frequencies double that of the mass-spring system demonstrated a sinusoidal loading pattern and phase plane patterns of the lower extremities resembling limaçons. Infants who oscillated at frequencies close to 1.5 times the natural frequency of the mass-spring system, however, exhibited slanted sinusoidal loading and demonstrated smooth circular phase plane plots. Infants who produced a mixed pattern had variable oscillating frequencies. Vallis (1998) used phase plane analyses to explain the differences and suggested that two attractor states were viable, that infants were not limited to the use of a single underlying movement pattern and that their adaptations to perturbations of the system may result in shifting from one attractor state to another attractor state. Less understood is whether the different phase

planes demonstrate a change in stability or whether they represent attractor states of variable stability.

From the infant bouncing studies, it appears that when infants are in stable attractors, they bounce at frequencies similar to model of a double spring system (Goldfield et al., 1993). Moderately skilled and skilled infants demonstrate two possible attractor states as two different bouncing frequencies were found (Vallis, 1998). These infants also demonstrate joint phase planes of particular shapes that represent clear spring-like motion (circle) or a combination of a spring with the addition of a forcing component (limaçon). Finally, preliminary evidence from one infant demonstrates that an infant can bounce at different frequencies when environmental conditions such as mass or number of springs are changed (Foo et al., 2001). Further research is required to investigate whether skilled infants can bounce under different conditions, whether these infants demonstrate various bouncing frequencies and whether phase planes can be used to describe the resultant action of the control mechanism at the joint. While previous studies have identified the outcomes of a limited number of environmental conditions and for a limited number of bounces, infant bouncing behaviour has yet to be systematically explored across multiple conditions.

The aim of this study was to determine whether infants produce similar or identical bouncing frequencies and kinematic behaviours in conditions with different spring frequencies. The study explored the bounce-to-bounce adaptations of timing and force regulation, bounce height and joint kinematics at the ankle, knee and hip joint, as well as intra- and inter-limb coordination as spring frequency increased. Lastly, visual inspection of bounce cycle phase plane patterns of bilateral ankle, knee and hip joints was used to characterize qualitatively the mechanical behaviours at each joint.

Objectives This study investigated infant bouncing behaviours and applied a dynamical systems approach to explain the findings. The ability of infants to adapt to environmental variations was analyzed with the following dependent variables:

- Bouncing frequency was compared to the natural frequency of the spring system;
- Bounce kinematics, including vertical height of each bounce and angular range of motion of the ankle, knee and hip joints;
- The amplitude of vertical ground reaction forces
- Regulation of timing through analysis of the percentage of time on the ground and the percentage of time to peak vertical ground reaction force.
- Stability of inter- and intra-limb coordination using angle-angle correlations
- Preferred stability of joint behaviour by means of phase planes, the relationship between angular position and velocity

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CHAPTER THREE

METHODS

GENERAL INFORMATION

Information regarding participants, experimental protocol, data acquisition, data reduction, data processing, statistical analyses and phase plane analyses are reported in detail in the following manuscripts. Instrumentation information, in addition to what is reported in the manuscripts, is provided in the current chapter.

Instrumentation

A spring system (SS) which was designed to simulate a standard baby bouncer, could be modified to change the natural frequency of the system. The natural frequency of the system was modified by the addition of springs and weights to approximate four spring frequency conditions (0.9, 1.15, 1.27 and 1.56 Hz); the average stiffness at each condition of the SS was 422, 836, 830 and 1251 N/m for conditions one to four (see Table 2). The SS was suspended from the ceiling over two AMTI force platforms. All kinematic data were recorded using 7 Vicon MX cameras.

Table 2 –Measured spring frequencies corresponding to the different spring stiffness' in each condition with the fixed mass.

Conditions	Total mass of the system (kg)	No. of springs	Spring stiffness (N/m)	Natural spring frequency (ω_n , rad/s)	Natural spring frequency (ω_n , Hz)
1	13.00	1	422	5.66	0.90
2	16.00	2	836	7.25	1.15
3	13.00	2	830	8.01	1.28
4	13.00	3	1251	9.80	1.56

CHAPTER FOUR

INFANT BOUNCING AT DIFFERENT SPRING FREQUENCIES

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Keyword: infant bouncing, ground reaction forces, skill adaptations, variability

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Conflict of Interest

No author has a personal or financial conflict of interest that could inappropriately influence their work.

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Abstract

Infants explore their environments through repetitive movements which are constrained or facilitated by the environmental context. Open-loop tasks become closed-looped when infants come in contact with an object or a surface, as seen in infant bouncing. The current study analyzed how typically developing infants bounced to four different system natural frequencies. Four pre-walking infants (age 9.7 months \pm 1.8) were placed in spring conditions with four natural spring frequencies of 0.9, 1.15, 1.27 and 1.56 Hz. All infants bounced above the natural spring frequency in all conditions suggesting that they do not solely behave like a mass-spring system. Two patterns of bouncing adaptations were identified. Three infants were found to be regulating bounce frequency, while one infant regulated the percentage of time on the ground. The vertical displacement of the trunk decreased across condition in infants who followed the task condition with increasing bounce frequency; however, it remained equal when percentage of time on the ground was regulated. Asymmetrical loading patterns and decreasing vertical ground reaction forces were found in all infants suggesting that a timing component is always regulated. The difference in bouncing pattern may be indicative of different bouncing skill level.

1. Introduction

During development, infants explore their environment and incorporate multiple degrees of freedom to create repetitive movements that are constrained or facilitated by the environmental context. Numerous studies have investigated motor acquisition of the lower extremity in open-loop tasks such as spontaneous kicking, where infants do not come in contact with a surface or other object. These studies have identified the differences in leg movement patterns in full-term and preterm infants (Heriza, 1988; 1991; Piek & Gasson, 1999), leg phase timing (Thelen, Bradshaw, & Ward, 1981) and postural effects (Jensen, Thelen, Ulrich, Schneider, & Zernicke, 1995; Jensen, Ulrich, Thelen, Schneider, & Zernicke, 1994) in infant kicking. It has been suggested that once a movement pattern has become stable through assembly and tuning, it can be transferred to contact task specific actions (Berthouze & Goldfield, 2008; Goldfield, Kay, & Warren, 1993; Thelen, 1995; Thelen, Kelso, & Fogel, 1987). Spontaneous supine kicking, for example, becomes closed-loop when infants come in contact with an object or surface (Angulo-Kinzler, Ulrich, & Thelen, 2002; Galloway & Thelen, 2004).

Infant bouncing development is a highly complex repetitive closed-looped and task-specific movement. Bouncing requires trunk and body weight support and the generation of a pushing force as the feet come in contact with the ground. Goldfield, Kay and Warren (1993) used a mass-spring model to explain infant bouncing. The researchers proposed that infants who bounced at the resonant frequency of the spring used their legs like springs-in-series and that peak bouncing frequency would match the resonant frequency of the spring. Matching a predetermined behaviour such as the resonant frequency of the system suggests that a preferred movement behavior or attractor state is achieved.

In the Goldfield et al. (1993) work, infant bouncing was recorded weekly for a minimum period of six weeks and the infant's peak bouncing frequency was compared to the predicted resonant frequency of a single and a double spring model. The researchers suggested that infants bounced close to the resonant period of the spring when they reached peak bouncing, defined as the bout with the greatest number of bounces. Infants bounced near 1.67 Hz corresponding to the predicted double spring model. In this model, the infant's legs were modeled to behave as a spring that matched the impedance of the external spring. The peak bounce frequency was clearly higher than the predicted single spring model, 1.25 Hz, which accounted for no leg stiffness. Similar infant bouncing frequencies of 1.63 Hz (Foo, Goldfield, Kay, & Warren, 2001) and 1.56 Hz (Vallis, 1998) were reported in subsequent studies.

Follow-up work by Goldfield and colleagues (with one infant) investigated skill acquisition to determine whether infants learned the low-dimensional dynamics of the task or whether a single pattern of bouncing was learned and applied under multiple conditions (Foo et al., 2001). Although data obtained were limited to one infant, the ability to generate bouncing behavior across different conditions was identified. In the baseline condition (one spring, no added weight), the peak bounce period corresponded to a bounce frequency of 1.63 Hz, similar to the frequency predicted with a double spring model (1.67 Hz). When the spring constant was increased, the infant bounced at 1.84 Hz, close to the frequencies predicted by both a single (1.72 Hz) and double (2.10 Hz) spring model (Foo et al., 2001). However, neither model predicted a bounce frequency close to the actual bounce frequency when a mass of 2.3 kg was added to the system. In this case, the infant bounced at 1.71 Hz while the single and double spring models predicted bounce frequencies of 1.12 and

1.49 Hz, respectively. These data suggest that there must be an active contribution to bouncing behavior that is beyond having the legs behave like springs in series.

Loading patterns from a load cell and phase plane plots of lower extremity data were used to qualitatively explore whether infants were using their legs as different mechanical systems (Vallis, 1998). Infants were characterized as ‘skilled’ bouncers when they produced multiple bounces in a row. The less skilled bouncers produced disorganized movement patterns with poor rhythm control resulting in sudden stops and starts throughout the bouncing bout. Moderately skilled and skilled bouncers showed two types of bouncing behaviors that were characterized by two distinct bouncing frequencies and two different load cell loading patterns. Infants that bounced at double the resonant frequency of the mass-spring system demonstrated a sinusoidal loading pattern of the spring, whereas infants who bounced closer to 1.5 times the natural frequency of the mass-spring system exhibited slanted sinusoidal loading patterns of the spring suggesting that more time was spent loading than unloading the spring. Vallis (1998) used phase plane analyses to explain the differences and suggested that two attractor states were viable, as two underlying mechanical parameters existed. The notion that two different strategies achieved the same movement behavior has also been seen in infant kicking.

Angulo-Kinzler, Ulrich and Thelen (2002) reported that 3-month-old infants used two strategies to move a mobile above their heads. Biaxial goniometers were placed on infant’s knees and once angular thresholds of 85° or 35° of knee flexion and extension were achieved, the mobile would rotate. Infants responded with two strategies where some infants moved with greater ranges of motion to surpass the threshold, while others sustained a position near the threshold to trigger the mobile with small fluctuations

(Angulo-Kinzler et al., 2002). Thus, infants as early as 3-months of age were able to adapt to task constraints and used two different strategies to achieve a task.

Providing that infants learn the low-dimensional dynamics of a motor task instead of a specific pattern, they should quickly adapt to changes in the task conditions. In bouncing, this would suggest that infants would not learn fixed leg stiffness or fixed timing of the forcing function (forcing frequency). Thus, successful bouncing when the systems' mass or spring stiffness is modified would require the infants to adapt to the new parameters with different forcing frequencies or leg stiffness (Foo et al., 2001; Goldfield et al., 1993).

The purpose of this study was to determine how infants generate bouncing in four different spring conditions. The current study differs from previous studies (Foo et al., 2001; Goldfield et al., 1993; Vallis, 1998), as infants were not observed longitudinally. Rather, infants were tested within a short developmental window and bouncing behavior on spring systems with different natural spring frequencies were recorded. In published literature, Goldfield et al. (1993) were the first to report individual bounce cycle data for bounce frequency and height; however, they only looked at a maximum of 12 continuous bounces. This study is the first to analyze individual bounce cycles with a large number of continuous bounce data. To understand the adaptations of behavior, we first asked whether infants bounced at the same frequency of the resonant frequency of the system. We hypothesized that the infants would produce bouncing frequencies close to the natural resonant frequency of the system across the different conditions suggesting that they are learning the low-dimensional dynamics of the task. We used vertical ground reaction forces (VGRFs) and vertical trunk displacement to characterize behaviour across the spring conditions.

2. Methods

2.1. Instrumentation

A spring system (SS) was designed to simulate a standard baby bouncer and could be modified to change the natural frequency of the system with the addition of extension springs and weights to approximate four spring frequency conditions (0.9, 1.15, 1.27 and 1.56 Hz). Average spring stiffness ranged from 422–1251 N/m. The SS was suspended from the ceiling over two AMTI force platforms. All kinematic data were recorded using 7 Vicon MX cameras.

2.2. Experimental Protocol

2.2.1. – Inert Mass (IM) Experiment

The first experiment was conducted to determine the natural oscillation frequency of a mass equivalent to the infant's mass. SS frequencies were measured experimentally with masses equivalent to each infant's mass. To reach the natural frequencies of the four spring conditions, the number of springs (up to three) and the amount of additional weight on the bar (up to 6 kg) were manipulated (Fig. 1a). For each condition, a mass was suspended from the SS and set into an oscillating motion. The oscillation of the mass was recorded using passive reflective markers attached to the system and mass. The oscillating frequency of the IM was calculated by counting the number of sinusoidal peaks divided by total time it took the SS to naturally damp out. Although the SS contained light damping properties, the damping ratio was considered negligible (1.35 to 5.49×10^{-3}).

2.2.2. – Infant Experiment

To complete the experimental protocol, each infant participated in two testing sessions within a 7-day period. At each session, the infant was weighed, dressed in a diaper, and placed in the spring system with the knees slightly bent and the balls of the feet

touching the force platforms (Vallis, 1998). Reflective markers were placed over body landmarks and at the top of the harness. Infants were given a few minutes to familiarize themselves in the 0.9 Hz spring condition (C1). Bouncing was then recorded for a period of up to 15 minutes in C1 followed by C2 (1.15 Hz), C3 (1.27 Hz) and C4 (1.56 Hz).

2.3. Participants

Nine typically developing pre-walking infants between 5 and 12 months of age were recruited for participation. Infants were included if they were able to support their head, sit on their own and demonstrated bouncing behaviour in a Jolly Jumper bouncer or Exersaucer at home. Infants were excluded if they walked independently or if they did not produce a minimum of 3–5 minutes of bouncing behaviour over the 15 minutes of continuous recording in the first condition of the study. Parents or guardians provided informed consent for the infant to participate. Only four of the nine infants tested produced bouncing behaviour and their data were included in the analysis.

2.4. Data Acquisition, Reduction and Processing

Three-dimensional kinematic data were recorded at 120 Hz (Vicon Nexus, Colorado) from reflective markers placed over the midsternum, 8th thoracic vertebrae and at the top of the harness (one on each bar of the apparatus). Ground reaction forces (GRF) were recorded at 600 Hz with a 10 N threshold using two Advanced Mechanical Technology, Inc. (AMTI, Model OR6-5) force platforms.

In each condition, bouncing bouts were defined as a minimum of six continuous bounces with no pauses. Experimental data were processed using Visual3D (C-Motion, Inc.). Vertical position of the 8th thoracic vertebrae (T8) or sternum was extracted and the linear velocity was calculated by taking the first derivative of the position. Analog data

from the two force platforms were low pass filtered (Butterworth 4th order cutoff of 10 Hz), summed and used to determine when the infant was in contact with the ground.

Post-hoc processing was conducted with MATLAB2008a. A bounce cycle was defined as the cycle from touch-down to subsequent touch-down using the GRF data. The mean and variability of the bounce frequency, T8 vertical displacement, peak vertical ground reaction force (VGRF), percentage of time on the ground, and percentage of time to peak force were analyzed.

2.5. Statistical Analyses

Statistical analyses were performed using SPSS 18.0 (SPSS Inc.). Independent sample *t*-tests were conducted for each infant to determine whether significant differences existed between the infant's bounce frequency and the oscillation frequency of the IM frequency of oscillation for each of the four conditions. Independent 4×4 (Infant × Condition) between-within factorial analyses of variance (ANOVAs) were conducted on all dependent variables. Post-hoc analysis consisted of one-way ANOVAs to identify significant differences between infants in each condition and repeated-measures ANOVAs to determine whether significant differences existed within each infant across conditions ($p \leq 0.01$). Pair-wise comparisons were analyzed to identify significant differences across all infants and all conditions, respectively. To avoid Type I error, *p*-values were corrected for the number of pair-wise post-hoc tests that were conducted ($p \leq 0.013$).

3. Results

Data were obtained from four typically developing infants (3 males, 1 female) aged 7.1 to 11.5 months ($M=9.7$, $SD=1.8$). The average masses and supine length of these infants were 10.67 (± 0.79) kg and 72.42 (± 3.12) cm, respectively. The infants spent between 140 and

315 minutes per week in a commercial bouncer (e.g., Jolly Jumper or Exersaucer) in their home. All infants completed the four conditions, except for infant 3 (I3) who did not complete condition 4 (C4). The number of bounces analyzed varied with 165, 427, 95 and 364 bounces analyzed for I1 through I4, respectively.

Significant interaction effects of Condition \times Infant ($p < 0.01$) and main effects of Condition ($p < 0.01$) and Infant ($p < 0.01$) were found for all dependent variables. Results of post-hoc pair-wise comparisons for each dependent variable are reported below and summary data presented in Table 1.

3.1. Bounce Frequency

In all conditions, infants bounced at significantly higher frequencies than the natural frequency of their corresponding IM ($p < 0.0005$). I2 and I4 bounced at similar frequencies and although significantly different than I1 ($p \leq 0.001$), the three infants appeared to follow the same pattern where the bounce frequency increased as the natural frequency of the system increased (see Fig. 2). Moreover, the difference between the infant bounce frequency and the natural system frequency was approximately constant for each infant (I1: 0.23-0.28 Hz; I2: 0.14-0.18 Hz; I4: 0.12-0.2 Hz). In these three infants, the variability in bounce frequency increased as spring frequency increased from C1 to C4. In contrast, I3 bounced at higher bounce frequencies than her IM frequencies in all three conditions ($p < 0.0005$) with no pattern identified across condition. In contrast to I1, I2 and I4, bounce frequency variability decreased as spring frequency increased for I3. Lastly, bounce frequency was significantly different across all conditions for each infant ($p < 0.0005$).

3.2. Trunk vertical displacement

Trunk (T8) vertical displacement varied significantly across all infants in all conditions except for I1 and I2 in C2 and C3, and I1 and I4 in C4 ($p < 0.0005$) (see Fig. 3a).

Specific patterns were identified across infants with T8 vertical displacement decreasing significantly from C1 through C4 in I1, I2 and I4 (all comparisons $p < 0.0005$). In contrast, in I3, although T8 vertical displacement was significantly higher in C2 than in C1 and C3 ($p < 0.0005$) the absolute mean difference in T8 vertical displacement was less than 3.0 cm. Although the variability of T8 displacement was low across all infants, variability decreased as spring frequency increased for I1, I2 and I4 while the variability increased and reached a plateau for I3.

3.3. Loading patterns on the ground

Peak VGRF: Figure 3b shows the mean peak VGRF values across the four conditions for each infant. Though I1 always produced the highest VGRFs, no Infant pair-wise comparisons similarities were found at any condition. However, two loading patterns were observed. The first pattern, seen in I1 ($p < 0.0005$) and I2 ($p \leq 0.002$), was characterized by a significant decrease across the four conditions although the decrease was less drastic in I2. Peak force variability also decreased as spring frequency increased for I1 and I2.

The second pattern, seen in I3 ($p < 0.0005$) and I4 ($p < 0.0005$), was characterized by a significant decrease in VGRF from C1 to C2 followed by minimal additional change. Peak force variability for I3 followed a similar pattern while for I4, the variability of peak force decreased from C1 to C2 followed by a slight increase.

Percent of time on the ground: Except for I3, no patterns emerged in the data and most pair-wise comparisons of data from the four conditions were significantly different for all infants (Fig. 3c). In I3, the percentage of time on the ground was not significantly different across any condition.

Specific comparisons showed that the percentage of time on the ground varied significantly by infant in C1 ($p < 0.0005$) and C4 ($p < 0.0005$) where percentage of time on

the ground was the lower in C1 than C4 for I1 and I2, higher in C1 than C4 for I3, and was equal in C1 and C4 for I4. In C2, all infants spent significantly different percentages of time on the ground ($p < 0.0005$), except for I1 and I4. In C3, I1 and I3 both had significantly higher percentage of time on the ground than I2 ($p < 0.0005$) and I4 ($p < 0.0005$). Percentage of time on the ground differed significantly between C1 and C2 ($p = 0.007$), C1 and C3 ($p < 0.0005$), C2 and C3 ($p < 0.0005$), and C2 and C4 ($p < 0.0005$) for I1, with percentage of time on the ground gradually decreasing then increasing across conditions. Except for C1 and C3, percentage of time on the ground was significantly different across all conditions for I2 ($p < 0.0005$) where the percentage time spent on the ground significantly decreased in C2 then gradually increased in C3 and C4. Except for C1 and C4, percentage of time on the ground was significantly different across conditions for I4 (all comparisons $p \leq 0.009$) with percentage of time on the ground increasing then decreasing across conditions. Percentage of time on the ground variability increased with spring frequency for I1, I2 and I4.

Time to peak force as a function of time on the ground: There were no similarities identified in either the percentage of time to peak force or the variability of percentage of time to peak force in the loading patterns amongst infants. Across conditions, only I1 showed no significant difference in the percentage of time to peak force (see Figure 3d).

Specific comparisons showed that the percentage of time to peak force differed significantly in C1 between I1 and I2 ($p < 0.0005$), I2 and I3 ($p < 0.0005$), I2 and I4 ($p < 0.0005$) with I2 loading to its peak force the quickest. In C2, percentage of time to peak force was significantly different between I1 and I2 ($p < 0.0005$), I1 and I3 ($p = 0.008$), I3 and I2 ($p < 0.0005$), and I3 and I4 ($p < 0.0005$) with I2 loading the fastest and I3 loading the slowest to peak force. With I2 reaching peak force the quickest in C3, I2 differed significantly from I1, I3 and I4 (all comparisons $p < 0.0005$). In C4, percentage of time to

peak force was significantly different between I1 and I2 ($p < 0.0005$) and I1 and I4 ($p = 0.011$), with I1 loading to peak force the fastest.

The variability of percentage of time to peak force increased from C1 to C2, but remained steady in C3 and C4 for I1. For I2, percentage of time to peak force differed significantly in all condition pair-wise comparisons with C4 ($p < 0.0005$) as I2 demonstrated equal loading and unloading in C4 in comparison to the first three conditions. The variability of percentage of time to peak force for I2 increased from C1 to C2, but decreased for C3 and then again for C4. Percentage of time to peak force differed significantly for I3 between C1 and C2 ($p < 0.0005$), but percentage of time to peak force variability gradually increased from C1 to C3 for I3 demonstrating similar loading approach. For I4, percentage to peak force differed significantly between C1 and C2 ($p < 0.0005$), C2 and C3 ($p < 0.0005$), C2 and C4 ($p < 0.005$), and C3 and C4 ($p = 0.003$), as C2 had the quickest loading than the other conditions. Percentage of time to peak force variability for I4 demonstrated the same pattern as I2, where it increased in C2 from C1, but decreased in C3 and C4.

In summary, I1, I2 and I4 bounced in similar increasing frequencies, matching the resonant frequency of the spring system, while I3's bounce frequency remained relatively constant across conditions. Trunk vertical displacement decreased across conditions for I1, I2 and I4, whereas I3 bounce height remained equal across conditions. VGRFs decreased for all infants across conditions, albeit at different rates. Percentage of time on the ground remained constant for I3, while the other infants demonstrated no patterns. Lastly, the time to peak force as a function of time on the ground did not change across conditions for I1, whereas other infants did not display any patterns.

4. Discussion

The objective of this study was to investigate if and how infants would adapt bouncing behavior to 4 different spring systems. All infants bounced at higher frequencies than the system's resonant frequency supporting the hypothesis that infants do not solely behave as a mass-spring system. These findings are in close agreement with Goldfield et al. (1993) original hypothesis.

Two patterns of adaptation were identified. The first pattern was observed in three infants and was characterized by increased bounce frequency that followed the task condition. These infants appeared to learn the low-dimensional dynamics of the task; however, evidence suggests that reaching the resonant frequency was not their only objective. These infants (I1/I2/I4) bounced above the resonant frequency of the spring with an additional constant, suggesting that they contributed an additional component while bouncing and maintained this consistent across conditions. This suggests that another variable, such as the VGRF, is being altered to maintain an equal constant above the resonant frequency. For the infants that followed the increasing spring frequencies, their bounce frequency variability also increased; however, the additional constant contribution of the infant remained the same. When bounce frequency is considered, one can suggest that infants who have already learned to bounce begin in a very stable attractor state and become less stable as reflected in the increased variability as the system natural frequency increased.

The second pattern was seen in only one of the four infants tested. The higher bounce frequency variability that I3 demonstrated in C1, in comparison to other infant's conditions, is indicative that this infant was not stably bouncing at 0.9 Hz and suggests that

another variable was controlled since the infant was still able to adapt to the changes in spring frequency.

As the spring frequency changed, infant's 1, 2 and 4 mean vertical trunk (T8) displacement and trunk variability decreased. Such results have not been explored in previous infant bouncing studies; however, similar findings in wrist oscillatory movements (Kay, Kelso, Saltzman, & Schöner, 1987) and limb coordination dynamics (Jeka & Kelso, 1995) have been reported. In the former study, wrist movement amplitude and peak velocity were investigated as wrist frequencies were manipulated at discrete increasing frequencies (1–6 Hz) of 1 Hz intervals. Results demonstrated that individual and coupled (left and right) wrists were able to match the auditory metronome and as wrist oscillation frequency increased, peak velocity increased and wrist movement amplitude decreased (Kay et al., 1987). In the latter study, when homologous arms and legs were coupled and subjects instructed to oscillate the limbs at increasing frequencies (1.0–2.25 Hz) by following an auditory metronome, results demonstrated that both the arm and leg followed the timing of the metronome very closely (Jeka & Kelso, 1995). As the task of the study (Jeka & Kelso, 1995) was for the coupled limbs to keep up with the auditory frequencies, the task drove the mean cycle amplitudes of the arms and legs to decrease as the oscillation frequency increased. Both aforementioned studies demonstrated that as the frequency of the task increased, movement amplitude decreased regardless if change in perturbation was conducted discretely or continuously. Thus, we can propose that additional changes to the system are secondary and are the result of regulating timing, the primary task.

Since not all the infants demonstrated a similar increasing bounce frequency pattern, an alternate infant bouncing strategy was considered. With a variable bouncing frequency across conditions and bounce frequency variability decreasing as spring

frequency increased, it was evident that I3's primary goal of the task was not to match or follow the resonant frequencies of the spring. Although differences were found in trunk vertical displacement across conditions, the differences were minimal less than 3 cm or 38% of the total vertical displacement of the trunk. The maximum T8 vertical displacements for I3 were less than all the other infants, indicating that the infant was not limited by the system and could have reached larger displacement; however, she maintained her vertical displacement relatively consistent. Interestingly, I3's percentage of time on the ground was not different across conditions suggesting that a timing component was being regulated, which may have resulted in the relatively fixed bounce height. We propose that the percentage of time the I3 remained in contact with ground demonstrated a second strategy to bouncing.

Regulatory strategies in infant bouncing have not previously been identified; nonetheless, the difference seen in bouncing behaviors, as Vallis (1998) highlighted, may be the result of bouncing skill level. The number of bounces produced by I3 was much lower than the other infants, which suggests, according to Vallis' method of skill characterization, that among this group of skilled infants, I3 was the least skilled. Although infants bounced using different regulatory methods, all infants demonstrated decreasing VGRFs and spent less time on the ground than in the air as spring frequency increased. We also notice the timing asymmetries correspond to steeper force loading than unloading rates, as the percentage of time to peak force is less than 44.9% for all infants in C1 to C3, but then varies (42.7–50.3%) across infants in C4.

The present study supports the earlier work showing that infant bouncing demonstrates two types of behaviour (Vallis, 1998), one where infants match the natural frequency of the system and the second, where infants regulate the percentage of time in

contact with the ground. The data underscore that infants used more than one type of adaptation in contrast to earlier suggestions by Goldfield et al. (1993).

The two adaptation strategies seen in the current study suggests that we can further investigate whether the two behaviors are able to be characterized through different phase planes, which would reflect distinct underlying mechanisms of behavior and correspond to changes of attractor states as Vallis (1998) suggested. Phase planes have been used to characterize the coordination of skilled musicians. Similar phase plane plots were found when opposing tasks and equal motor goals were instructed, where movement trajectories were closely related to timing accuracy (Balasubramaniam, Wing, & Daffertshofer, 2004). We believed that asymmetrical loading seen in infant bouncing is related to timing accuracy.

Balasubramaniam, Wing and Daffertshofer (2004), instructed skilled musicians to oscillate their index finger without surface contact at different frequencies (1.00, 1.33 and 2.00 Hz) in different paired task (synchronize and syncopate) and motor (flexion and extension) goals. Oscillating trajectories were analyzed in four conditions: self-paced, synchronize—when peak flexion (fON) or peak extension (eON) occurred on the beat, and syncopate—when peak flexion occurred off the beat (fOFF). Both synchronization conditions such as flexing and extending on the beat encompass equal task goals, however, they are not equivalent. Through visual inspection of phase planes with an auditory metronome, flexion off the beat syncopation is equivalent to flexion on synchronization albeit at 180° phase shift. In addition, movement asymmetries were found in all paced condition, but movement trajectories became more symmetrical with increasing frequencies, as seen in the unpaced condition. In comparison to unpaced conditions, subjects moved at higher velocities in shorter time spans towards the temporal target (fON,

fOff, eON) and produced slower movements in the return phase. The researchers proposed that asymmetry was related to reducing timing errors, as the greater asymmetry found, the closer it was to synchronous timing in such way that when the finger arrived early and produced large asynchrony, the finger would compensate by a longer return phase that would correct for the asynchrony (Balasubramaniam et al., 2004).

Although infants did not adapt to the changes in spring frequencies using the same strategy, the concept that asymmetry is a result of timing regulation can still be applied to all infants as both strategies were regulating a timing component. Thus, the asymmetrical loading VGRFs produced by the infant when s/he came in contact with the ground was a result of how infants regulated and corrected their timing component (bounce frequency or percentage of time on the ground). Further analysis is needed to see whether the infant's adaptation to the spring frequencies are represented differently as a function of phase planes, as Vallis (1998) suggested.

The negative relationship between bounce frequency and the T8 vertical displacement in three of the infants raises the question of whether the mechanical instrumentation (the springs) of the SS restricted the infant's ability to bounce at higher amplitudes. For three infants, as spring frequency increased, the infant's bounce frequency also increased to keep pace with the natural frequency of the system. Although the spring(s) at times collapsed as the infants pushed upwards, their momentum was never restricted due to a fixed position since the SS hung from a collapsible chain. The infants could have squatted lower to pull the extension spring additionally against the spring's resistance; however, that may have resulted in working against the natural frequency of the system. I4 bounced at higher amplitudes than the other infants. Since all infants bounced from the same SS this suggests that the heights reached by I1, I2 and I3 were not restricted

by the mechanical components of the SS but rather another variable. As infants were not restricted in bounce height, infants were either driven to match the natural frequency of the SS to regulate their own bounce frequency or driven to regulate the percentage of time on the ground.

A common pattern seen in all infants is the decreasing peak VGRFs produced. As bounce frequency increased the infants' peak VGRFs and variability decreased. Peak force has not yet been examined in the context of infant bouncing, but one can turn to jumping studies to find possible explanations. As in infant bouncing, jumping requires the legs to flex and extend; however, jumping requires the propulsion of the whole body from a surface with either one or both feet and the ability to land and balance on two feet when coming in contact with the ground. In an earlier study, the effects of different heights (40, 60 and 100 cm) on impact forces during landings were investigated in 3 males using different regression models (Dufek & Bates, 1990). For the purpose of this study, the regression models will not be discussed but instead results from jump height differences will be incorporated. Dufek and Bates (1990) found that as height increased, peak force increased. The linear relationship between height and peak force is attributed to the higher gravitational forces placed on the body when jumping from a higher distance. The directly proportional relationship between height and VGRF is also representative of a mechanical system. The relationship between the T8 vertical displacement and the VGRFs, however, is curvilinear for all infants, as VGRFs decrease in C2 from C1 and then plateau at higher frequencies. Therefore, the non-linear relationship between the T8 vertical displacement and the VGRFs further supports the concept that infants bouncing behaviour is not a replicated model of a mass-spring system or a pure mechanical system.

The current study's results of increasing bounce frequencies found in three infants resemble the findings from Angulo-Kinzler, Ulrich and Thelen (2002). In the current study, three infants produced very similar increasing bounce frequencies, but when their feet came in contact with the ground it was different in how each infant loaded and unloaded their forces and how much time they spent on the ground. Like the 3-month-old infants who were able to move a mobile by reaching a desired angular threshold of knee flexion and extension in two different methods (Angulo-Kinzler et al., 2002), I1, I2 and I4 bounced at similar increasing frequencies; however, the percentage of time each infant was in contact with the ground varied and the method of loading and unloading their forces were exclusive to each infant.

In light of our findings, our data strongly indicates that infant bouncing behavior is not limited to one method of adaptation. We suggest that the ability to adapt to the frequency changes of a spring is dependent on the ability to manipulate multiple variables when performing a repetitive task. The differences between the two strategies of adaptation may be related to the control of degrees of freedom of multiple joints in the lower extremity (Sporns & Edelman, 1993), which in turn may demonstrate a difference in bouncing skill. Regulating bounce frequency (timing accuracy of each bounce cycle) is a highly complex task, which requires altering variables, such as joint angles and VGRFs when you come in contact with the ground. Alternatively, regulating the percentage of time on the ground demonstrated limited range of trunk displacement and resulted in altering the same variables but with smaller joint ranges than the other infants. Consequently, both strategies require a timing component to be regulated; however, the differences between the strategies are over which degrees of freedom the infant has the most amount of control.

5. Conclusion

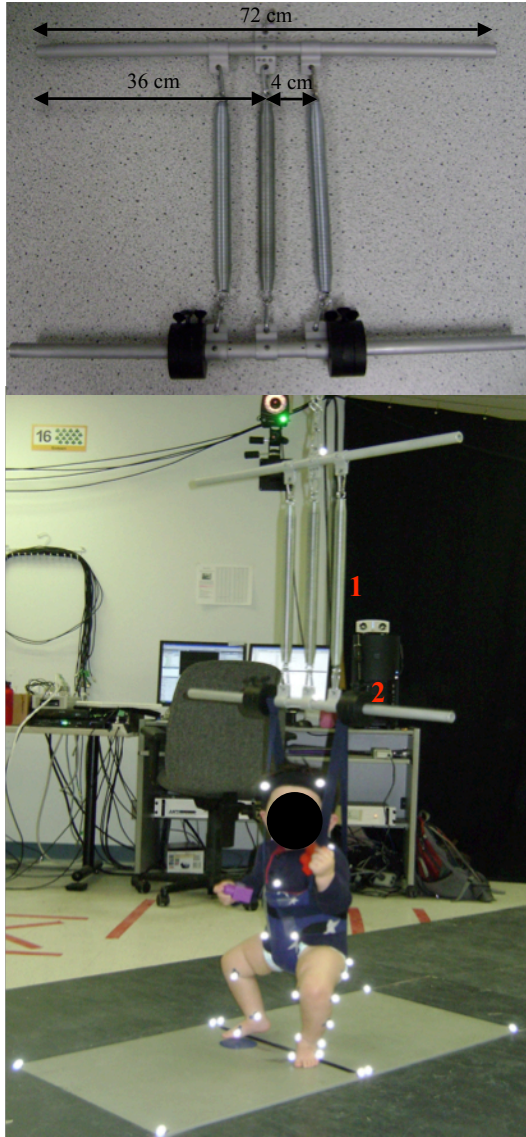
The present study demonstrated the adaptation behaviors of four infants, three who bounced with a similar increasing bounce frequency strategy and one with a different bounce frequency outcome. We have identified that infants do not bounce like a mass-spring system as they always bounced above the natural frequency of the SS. As all the infants were able to adapt to the changes of the SS, the study partially corroborates Goldfield et al. (1993) original hypothesis that once infants have learned a skill, they are able to adapt to the low-dimensional dynamics of the task rapidly. Their hypothesis, however, was limited, as it did not include different strategies of adaptation as we have shown. Three infants adapted to the new parameters of the SS and changed their forcing frequency and leg stiffness (Foo et al., 2001; Goldfield et al., 1993), but one infant was still able to adapt to the changes with more rigid leg stiffness and fixed timing of the frequency. Our findings are closely related to the previous work by Vallis (1998) where two types of bouncing behaviours were detected. In addition to supporting Vallis' initial findings, we demonstrated that skilled infants are regulating a timing component, which they have the most amount of control over when they come in contact with the ground. The regulated timing component was either bounce frequency or percentage of time on the ground. Joint mechanisms of control (phases-plane) analyses are still required to further support Vallis' explanation of why the infants are categorized in two strategy groups. Their underlying mechanism of control would clarify whether strategy of bouncing can demonstrate a shift of one attractor state to another attractor state, or whether they remain in the same attractor.

Table 1 – Inert Mass (IM) values and mean and standard deviation for each participant for all dependent variables summarized.

Infant Condition	I1				I2				I3			I4			
	1	2	3	4	1	2	3	4	1	2	3	1	2	3	4
IM Frequency (Hz)	0.87	1.16	1.27	1.55	0.91	1.14	1.28	1.57	0.92	1.18	1.29	0.90	1.14	1.26	1.56
Bounce Frequency (Hz)	1.14 ±0.08	1.39 ±0.23	1.52 ±0.17	1.83 ±0.22	1.05 ±0.07	1.29 ±0.13	1.42 ±0.13	1.75 ±0.19	1.98 ±0.38	1.55 ±0.24	1.79 ±0.23	1.07 ±0.05	1.26 ±0.13	1.42 ±0.15	1.76 ±0.18
T8 Vertical Displacement (m)	0.24 ±0.07	0.18 ±0.07	0.14 ±0.05	0.11 ±0.02	0.26 ±0.03	0.19 ±0.04	0.14 ±0.03	0.10 ±0.01	0.05 ±0.02	0.08 ±0.03	0.06 ±0.03	0.39 ±0.08	0.26 ±0.04	0.17 ±0.04	0.11 ±0.02
Peak VGRF (N)	84.1 ±40.7	56.1 ±22.5	42.0 ±19.5	33.3 ±12.3	31.4 ±11.1	29.0 ±10.1	26.5 ±8.9	23.1 ±7.5	80.4 ±31.0	39.3 ±18.3	40.9 ±26.1	76.4 ±27.6	33.2 ±11.0	29.1 ±13.5	29.5 ±18.2
Percent of time on ground (%)	36.6 ±5.5	34.6 ±5.8	40.1 ±9.4	39.4 ±11.1	26.4 ±8.7	23.9 ±9.2	27.1 ±9.7	34.7 ±11.5	45.8 ±9.8	44.0 ±15.3	42.7 ±12.0	29.8 ±4.2	34.3 ±8.4	26.7 ±9.2	29.2 ±10.8
Percent of time to peak force (%)	43.6 ±12.3	42.6 ±19.4	44.0 ±18.2	42.7 ±20.9	34.3 ±22.2	34.5 ±23.0	29.9 ±22.7	50.3 ±20.4	42.4 ±12.1	49.4 ±14.0	44.9 ±16.4	44.6 ±10.4	35.8 ±26.1	42.5 ±23.2	48.1 ±18.4

FIGURES

A



B

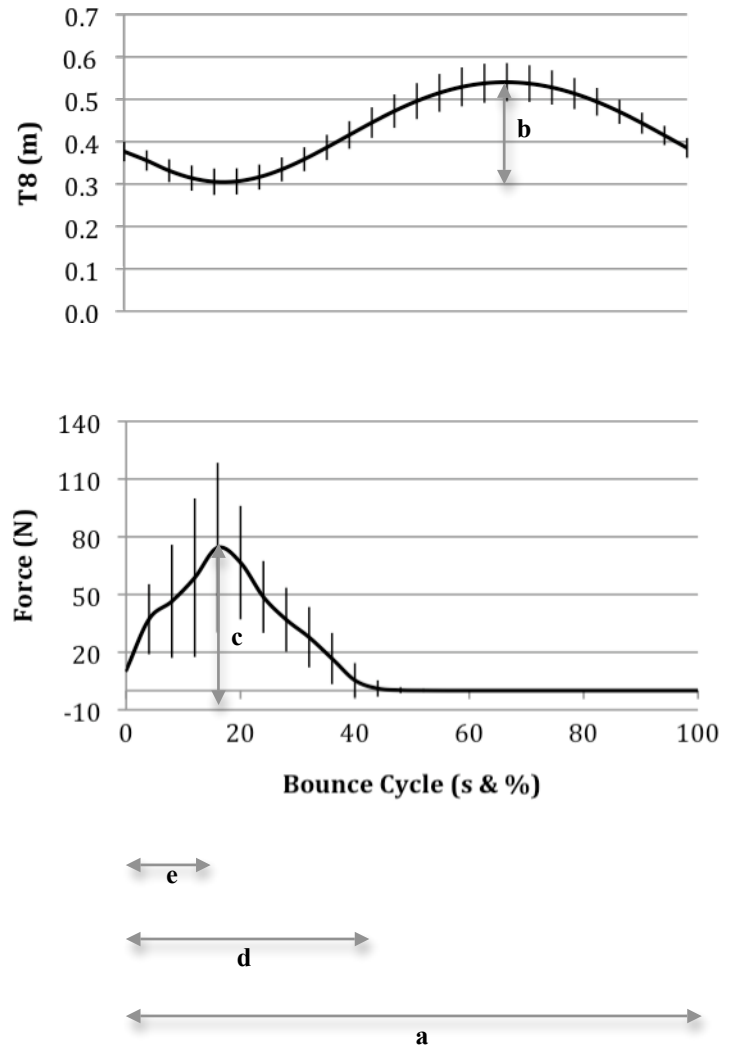


Figure 1A - Spring system apparatus with three springs (1) and 3 kg disc weights (2) and dimensions. Infant instrumented with full marker set. 1B – Experimental VGRF and T8 displacement data demonstrating the five (arrows) dependent variables in the study: a) bounce frequency; b) T8 displacement; c) peak force; d) percent of time on the ground; and e) percent of time to peak force.

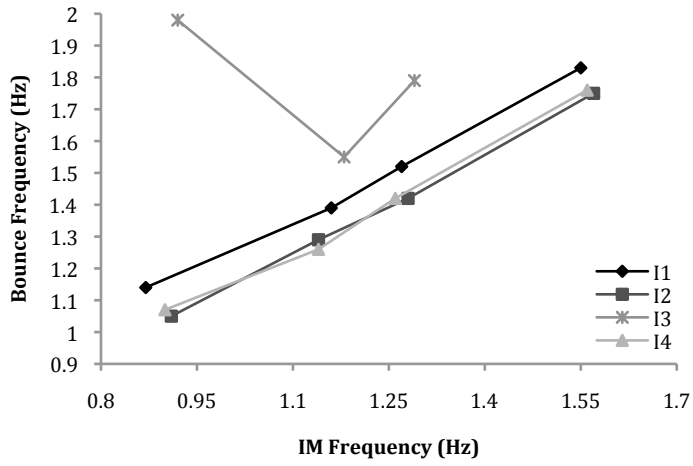


Figure 2 – Performed bounce frequency versus set inert mass (IM) frequency for all infants. I1-I4 denotes infant 1 through 4.

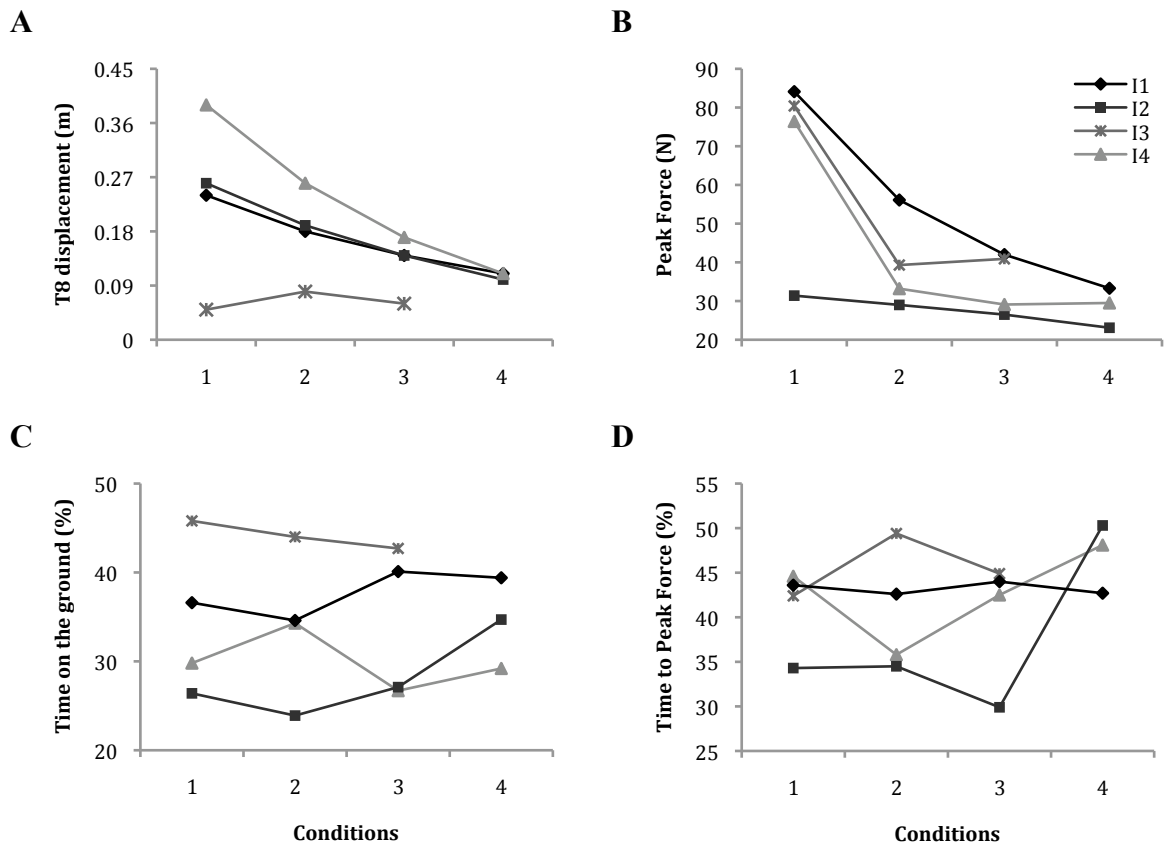


Figure 3 –T8 vertical displacement (A), peak force (B), percentage of time on the ground (C) and percentage of time to peak force (D) for C1-C4 for all infants.

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CHAPTER FIVE

INFANT BOUNCING: LEG MOVEMENTS AND MECHANISMS OF CONTROL

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Keyword: infant bouncing, phase planes, correlations, range of motion, variability

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Conflict of Interest

No author has a personal or financial conflict of interest that could inappropriately influence their work.

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Abstract

Within the literature of infant development, infant bouncing, a closed-loop task-specific movement has been the least studied. The current study analyzed cycle-to-cycle control mechanisms and leg movements of skilled infant bouncers during four different spring conditions. Four pre-walking infants (age 9.7 months \pm 1.8) were placed in spring conditions with four natural spring frequencies of 0.9, 1.15, 1.27 and 1.56 Hz. Infants responded to the frequency modifications of the spring system using one of two timing strategies. When infants matched their bounce frequency to the natural frequency of the system their joints ranges of motion decreased and they demonstrated a shift from non-spring-like to circular spring-like phase planes. Moderate to high correlations were found for inter- and intra-limb coordination. Conversely, when an infant regulated time on the ground, joint range of motion did not change across conditions and inter- and intra-limb correlations were low to moderate; phase planes remained circular spring-like for this infant. We found that the amount of deviation from a spring-like circular phase plane was attributed the joints ranges of motion when infants are on the ground. The differences in strategy represent a distinction in bouncing skill level.

1. Introduction

Much of the literature in infant development has focused on understanding spontaneous and task specific movements such as leg kicking, reaching and early locomotion. A common theoretical approach used to investigate such motions is dynamical systems theory, which addresses how nonlinear systems self-organize when a control parameter of that behaviour is altered (Kamm, Thelen, & Jensen, 1990; Kelso & Schöner, 1988; Shumway-Cook & Woollacott, 2001, p 18; Thelen, 1995). Dynamical systems theory has been primarily applied to coordination dynamics (Jeka & Kelso, 1995; Jeka, Kelso, & Kiemel, 1993; Kay, Kelso, Saltzman, & Schöner, 1987; Kelso & Jeka, 1992; Kelso, 1984; Kelso & Schöner, 1988; Treffner & Turvey, 1995) and motor development (Adolph, Vereijken, & Shrout, 2003; Goldfield, 2007; Goldfield, Kay, & Warren, 1993; Heriza, 1988a; 1988b; 1991; Holt, Saltzman, Ho, Kubo, & Ulrich, 2006; Kamm, Thelen, & Jensen, 1990; Newell, Liu, & Mayer-Kress, 2003; Piek & Gasson, 1999; Schöner & Thelen, 2006; Smith & Thelen, 2003; Thelen, 1985; 1994; 1995; 2005; Thelen, Fisher, & Ridley-Johnson, 1984; Thelen, Schöner, Scheier, & Smith, 2001; Thelen, Skala, & Kelso, 1987). Within this theoretical framework, one task-specific closed-looped movement, infant bouncing, is less understood.

Goldfield, Kay and Warren (1993) previously investigated infant bouncing with infants who bounced while hanging from a single spring. They applied a forced linear mass-spring model to explain the data from a dynamical systems perspective. The researchers proposed that through assembling and tuning a movement's low-dimensional dynamics, spontaneous activity becomes a "task-specific action pattern" (Goldfield et al., 1993, p 1129). Assembly involves the self-organization of a number of subsystems, where series of evolving and dissolving patterns of varying dynamical stability known as attractor states are

produced. Once movements are assembled, they refine (tune) to adapt to the demands of a task and the environmental conditions of the system (Goldfield et al., 1993; Heriza, 1991; Smith & Thelen, 2003; Thelen, 1995). Over time, movement patterns become stable and then are able to adapt to new environmental conditions.

In earlier findings from our laboratory, infants bounced at frequencies significantly higher than the natural frequency of the spring system (Habib Perez et al., unpublished). Three of our four infants increased their bounce frequency as the natural frequency of the spring system was increased. These three infants spent less time on the ground than in the air and the percentage of time on the ground varied across these infants. In contrast, one infant appeared to regulate the amount of time on the ground regardless of spring condition. All infants produced decreasing vertical ground reaction forces (VGRF) as the natural frequency of the system increased. Further analysis indicated VGRF loading and unloading were predominantly asymmetrical. The technique in which infants loaded and unloaded their VGRFs was related to the oscillatory movements of bouncing behavior and associated to a timing goal.

Based on findings from another study (Balasubramaniam, Wing, & Daffertshofer, 2004), we suggested that the asymmetries found in the VGRF loading and unloading patterns were attributed to timing error correction. In a study by Balasubramaniam, Wing and Daffertshofer (2004), skilled musicians demonstrated movement asymmetries when they were asked to synchronize and syncopate their index finger without surface contact at different frequencies (1, 1.33 and 2 Hz). Movement trajectories became more symmetrical with increasing frequencies, as seen in the self-paced condition. In comparison to self-paced conditions, subjects completed faster movements in shorter time spans towards the temporal target and slower movements in the return phase in all paced conditions. These researchers

proposed that the deviation from symmetry was related to the reduction in timing errors, as with greater movement asymmetry the closer the finger was to synchronous timing. The finger arrived early and produced large asynchrony, and compensated by a longer return phase that corrected for the asynchrony (Balasubramaniam et al., 2004). As infant bouncing is an oscillatory behaviour, we proposed that infants demonstrated one of two strategies of bouncing where a timing objective was always being regulated.

In the first strategy, infants' appeared to manipulate different movement characteristics such as VGRFs to regulate their bounce frequency at a fixed constant of approximately 0.2 Hz above the natural frequency of the spring system. As the natural frequency of the system increased, the vertical trunk displacement decreased. Similar findings have been seen in studies that manipulated wrist and limb oscillations frequencies (Jeka & Kelso, 1995; Kay et al., 1987). In these studies, the primary goal was to match the frequency of an auditory metronome. Movement amplitudes decreased with increasing frequency regardless of whether the changes in the driving frequency were discrete or continuous.

The infant using the second strategy demonstrated variable bounce frequencies suggesting that this was not the regulated variable. Rather, this infant appeared to be manipulating the VGRFs to achieve a consistent percentage of time on the ground, which resulted in a relatively consistent vertical trunk displacement across all spring conditions. We proposed that changes in vertical trunk amplitude are a result of regulating timing.

Our results suggest that infants adapt to changes of the spring frequencies in two manners; however, the underlying leg joint mechanics used when infants' come in contact with the ground as they regulate their timing component remain unclear. Indirect biomechanical analyses, such as inverse dynamics, use ground reaction forces on rigid

bodies to calculate net moments of force of individual joints. This method provides the most accurate approximation to the relative effort produced by the joints (Whittlesey & Robertson, 2004). However, for this method to be applied, ground reaction forces from multiple trials of single foot contact with the ground are required. When these data are not possible to obtain, graphical representations of joint kinematics such as phase planes and angle-angle plots provide insight to joint control mechanism and intra- and inter-segmental coordination.

Visual representations of joint kinematics, specifically phase plane plots, have been used to demonstrate preferred movements patterns. Stable movement patterns or attractor states occupy a preferred space in the plot (Heriza, 1991). The kinematic organization of leg movements has been investigated across various studies. For example, phase plots have been used to examine the differences between leg movements in normal and preterm infants (Heriza, 1988a; 1988b) and in infants with white matter disorder (Fetters, Chen, Jonsdottir, & Tronick, 2004), to differentiate finger syncopation and synchronization in skilled musicians (Balasubramaniam et al., 2004), to compare locomotion in normal adults to those with hemiparetic neurological impairment (Winstein & Garfinkle, 1989), and to explain different infant bouncing behaviors (Vallis, 1998). Circular phase planes illustrate a mechanical spring-like motion where there is a constant relationship between velocity and position. Other joint patterns take the shape of limaçons, where an embedded loop or a cusp exists within a circular phase plane. These deviations from a circular shape demonstrate a movement reversal or a sudden cessation of forces and movement interruption (Winstein & Garfinkle). In infant bouncing, circular phase planes at the ankle, knee and hip suggest that the leg behaves like a spring.

Earlier work in infant bouncing suggested that those infants who matched the resonant frequency of the system used their legs like springs. Specifically, Goldfield and

colleagues (1993) used a forced linear mass-spring model to explain bouncing. The best predictions for bouncing were obtained when the model accounted for the infant's legs to behave like springs that matched the impedance of the external spring (Goldfield et al., 1993). A circular phase plane with an embedded loop or cusp suggests that the joints are working as springs in series with the addition of a forcing function (Vallis, 1998) and although not analyzed, it is likely that the timing of the embedded loop or cusp matches the timing application of force to the ground. Limitations of previous studies are that visual inspections of phase plots have been categorized with multiple cycles overlapping, which although providing a general idea of the pattern, conceals cycle-to-cycle variations. In the current study, visual inspection of individual bounce cycle phase planes from each joint provided a better understanding of the control mechanisms infants use to bounce under varying conditions.

Environmental changes also influence the underlying inter- and intra-limb coordination used to generate specific movements. One method that can be used to illustrate preferred coordination patterns are angle-angle relationships. Angle-angle correlations have been used to identify interjoint coordination of the ankle, knee and hip during air kicking when three-month-old infants were placed in three postural positions—supine, angled, and vertical (Jensen, Ulrich, Thelen, Schneider, & Zernicke, 1994). There was a postural effect with the hip joint range of motion (ROM) decreasing as posture became more upright while the knee and ankle joints ROM remained the same in all three positions. Hip-knee intra-limb coordination was affected by posture, as infants demonstrated the highest intra-limb synchrony in the vertical position ($r \approx 0.87$) followed by the supine ($r \approx 0.74$) and angled ($r \approx 0.45$) positions (Jensen et al., 1994). Gravitational demands appeared to influence infant

kicking by restricting the range of motion of the kicks and increasing the intra-limb synchrony between the hip and knee joints.

The purpose of this study was to investigate the mechanism of control used by skilled infant bouncers to bounce in four different spring conditions. The current study differs from previous studies (Foo, Goldfield, Kay, & Warren, 2001; Goldfield, Kay, & Warren, 1993; Vallis, 1998), because the kinematics of joint movements for individual bounce cycles were analyzed over four spring conditions. To understand the adaptations of infant bouncing behavior to the four spring environments, we analyzed the differences in joint ROM, joint inter- and intra-limb coordination and the probability of using different movement patterns as reflected through visual analysis of phase planes. We hypothesized that lower extremity joints' ROM would decrease as the natural frequency of the spring system increased. We also hypothesized that as the natural frequency of the system increased, infants would shift towards a more circular phase plane pattern suggesting that their limbs become increasingly spring-like. Preliminary data were presented at the Canadian Society for Motor Learning and Sports Psychology (Habib Perez, Robertson, Baddour & Sveistrup, 2010) and the Society for Neuroscience (Habib Perez et al., 2010). Comprehensive methods are reported in Habib Perez et al. (unpublished).

2. Methods

A spring system (SS) was designed to simulate a standard baby bouncer. The natural frequency of the system was modified with the addition of springs and weights to approximate four conditions (0.9, 1.15, 1.27 and 1.56 Hz); the average stiffness at each condition of the SS was 422, 836, 830 and 1251 N/m for conditions one to four. The SS was

suspended from the ceiling such that the infant bounced on two AMTI force platforms. All kinematic data were recorded using 7 Vicon MX cameras.

The natural frequency of the SS was measured with masses equivalent to each infant's mass for each spring condition. The four experimental conditions were established by manipulating the system's total mass (infant, bar, harness mass) and the number of springs (up to three).

Each infant participated in two testing sessions within a 7-day period. At each session, the infant was weighed, dressed in a diaper, and placed in the spring system. Initial position in the SS was with the knees slightly bent and the balls of the feet touching the force platforms (Goldfield et al., 1993; Vallis, 1998). Reflective markers were placed over body landmarks and at the top of the harness. Infants were given a few minutes to familiarize themselves with the 0.9 Hz spring condition (C1). Bouncing was then recorded for a period of up to 15 minutes in C1 followed by C2 (1.15 Hz), C3 (1.28 Hz) and C4 (1.56 Hz).

Infants were included if they were able to support their head, sit on their own and demonstrated bouncing behaviour in a Jolly Jumper or Exersaucer at home. Infants were excluded if they walked independently or if they did not produce a minimum of three to five minutes of bouncing behaviour within the 15 minutes of continuous recording in the first condition of the study. Parents or guardians of the infants provided informed consent for the infant to participate. Four of the nine infants recruited produced bouncing behavior and their data were included in the analysis.

Three-dimensional kinematic data were recorded at 120 Hz (Vicon Nexus, Colorado) from reflective markers placed bilaterally over the first metatarsals, fifth metatarsals, medial and lateral malleoli, shanks, medial and lateral femoral epicondyles, thighs, greater trochanters, anterior and posterior superior iliac spines (ASIS and PSIS), and acromion

processes. Additional markers were placed over the midsternum, 8th thoracic vertebrae, on four locations around the head, and at the top of the harness. Ground reaction forces (GRF) were recorded at 600 Hz with a 10 N threshold using two Advanced Mechanical Technology, Inc. (AMTI, Model OR6-5) force platforms.

Data were analyzed for periods when the infant produced a minimum of six continuous bounces with no pauses. Kinematic data were processed using Visual3D (C-Motion, Inc.). Movement trajectories were low-pass filtered (Butterworth 2nd order cutoff frequency = 3 Hz). Flexion and extension angles of the ankle, knee and hip joints were computed by calculating the angle between two vectors, which were created using the spatial coordinates of three selected markers. Angular velocities were calculated as the first derivative of the joint angle positions. The vertical position of the 8th-thoracic vertebrae (T8) was extracted. Analog data from the two force platforms were summed and analyzed to determine when the infant was in contact with the ground. Analog force data were low-pass filtered (Butterworth 4th order cutoff frequency = 10 Hz).

Post-hoc processing was conducted with MATLAB2008a. A bounce cycle was defined using the VGRF data as the cycle from touch-down to touch-down using the summed VGRF data. The mean and variability of the intra-limb correlations (ankle-knee, knee-hip, ankle-hip), inter-limb correlations (left and right ankle, knee and hip), and the magnitudes of ankle, knee and hip ranges of motion of the left and right legs were analyzed. The time base of each bounce cycle was normalized to 100% for the ankle, knee and hip joint angle data, the T8 position and the VGRF data. Normalized data were used for visual representation.

Phase plane plots (joint angular position plotted against joint angular velocity) were used to illustrate joint movement patterns. Individual phase plane plots for each joint were

qualitatively coded using a system of seven shapes. A polar coordinate system was used and the amplitude of each phase plane was normalized. The coding method used to describe the mechanisms of control included a circle, a spiral, a convex, dimple, cardioid or trisectrix limaçon, and other (see Table 1). A circle represents the phase plane seen in a spring system. Spirals denote decreasing circular amplitudes, comparable to a damped spring-mass system. Other shapes, from convexes to trisectrices indicate an increasing deviation from a simple spring and reflect the addition of different amounts of force added to the joint. Two coders initially analyzed 990 phase planes with less than 20% interpretation difference. A single coder analyzed the remaining 3,214 phase planes. If two shapes occurred within the same phase plane (i.e., a trisectrix followed by a dimple), the first shape was coded. The data for the left and right joints were combined when inter-limb correlation coefficients were of moderate to high values (0.7 and above) (Vincent, 2005, p 103). Frequency counts of the shapes observed for all the joints were conducted to describe the underlying mechanism of control and to describe the change in the mechanism of control as the spring condition changed.

Pearson correlation coefficients (r) were used to determine intra- and inter-limb coordination, where each correlation indicated the relationship between movements at one joint with respect to another joint (Winstein & Garfinkle, 1989). Each joint of the lower extremity was coupled with another joint of the same extremity (intra-limb) and their equivalent joint on the contralateral leg (inter-limb). The relationships analyzed were ankle-knee, ankle-hip, and knee-hip for each leg, as well as the right and left sides for the ankle-ankle, knee-knee, and hip-hip angle-angle correlations. The mean and standard deviation r -values were calculated for each condition and for each infant separately. A Pearson

correlation of +1.0 and -1.0 represented coordinated movements of two joints in phase or 180 degrees out-of-phase, respectively.

Statistical analyses were performed using SPSS 18.0 (SPSS Inc.). Paired-sample *t*-tests were conducted for each infant to determine whether the left and right ankle, knee and hip ranges of motion were significantly different for all conditions. Since the majority of the left and right lower extremity joint ranges of motion were significantly different through all conditions, six independent repeated-measures analyses of variance (ANOVAs) were conducted for each infant to identify significant differences across conditions for the left and right ankle, knee and hip ranges of motion. When significant main effects of Condition were found, pair-wise comparisons were used to identify exact differences between all conditions ($p < 0.05$). In addition, nine independent repeated-measures ANOVAs were conducted for each infant on intra-limb correlations (ankle-knee, knee-hip and ankle-hip of the left and right leg) and inter-limb correlations (left and right ankle, knee and hip joints) to determine whether significant differences existed within each infant across conditions. A significance of $p < 0.05$ was set in pair-wise comparisons.

3. Results

Data from four typically developing infants (3 males, 1 female) aged 7.1 to 11.5 months ($M=9.7$, $SD=1.8$) were included in the analysis. The average masses and supine lengths of the infants were 10.67 (± 0.79) kg and 72.42 (± 3.12) cm, respectively. All infants completed the four conditions, except for infant 3 (I3) who did not complete condition 4 (C4). An equal number of bounces in each condition were analyzed for each infant (Infant 1 (I1): 165; I2: 427; I3: 95; and I4: 364).

3.1. Joint kinematics and Inter-limb Coordination for the Ankle, Knee and Hip

The mean ranges of motion for the left and right ankles, knees and hips for all infants are illustrated in Figure 1. In general, ranges of motion gradually decreased as the frequency of the spring system increased for I1, I2 and I4. Ranges for I3 remained constant across conditions. To illustrate the kinematic and kinetic data of the three infants who produced similar bounce strategies, I4's means and standard deviations for the ensemble-averaged and time-normalized T8 vertical displacements and all left and right joint ranges of motion, as well as the VGRFs are displayed in Figure 2A-D. The kinematic data follow synchronous sinusoidal patterns and all minima occur when the infant is on the ground near the peak VGRF. Similar findings were recorded for I3; however, the joints ranges of motion did not decrease across conditions. Figure 3A-C displays the kinematic and force data for I3.

Significant differences in paired *t*-tests were found between the left and right ankle, knee and hip ranges of motion for all four conditions for I2 ($p \leq 0.001$) and I4 ($p < 0.0005$). I1 demonstrated significant differences across all conditions for left and right ankle ($p \leq 0.007$), knee ($p < 0.0005$) and hip ($p \leq 0.001$) ranges of motion except for in C1 and C2 left and right ankle ranges of motion, and C2 and C3 left and right hip ranges of motion were not significantly different ($p \geq 0.059$). I3 demonstrated significant differences across all conditions in left and right ankle ($p \leq 0.004$), knee ($p \leq 0.001$) and hip ($p \leq 0.013$) ranges of motion except for in C2 left and right ankle and knee ranges of motion, and C3 left and right hip ranges of motion were not significantly different ($p \geq 0.08$).

Table 2 displays the means and standard deviations for inter-limb correlation coefficients for all three joint pairs (left vs. right) and all infants. *p*-values for the pair-wise comparisons between conditions for each infant's range of motion and inter-limb correlations are reported in Table 3 with significant levels marked by bold text and asterisks,

respectively. Overall, I1, I2 and I4 demonstrated moderate to high inter-limb correlation coefficient values, while I3 displayed low to moderate correlation coefficient values. High, moderate and low correlation coefficient values are denoted in bold, regular and grey text. Note that, the majority of inter-limb correlations for I1 were moderate with high correlations (.72 to .92) in C1. All inter-limb correlations for I2 and I4 were high, with a small range from .89 to .96 and .81 to .96, respectively. I3 demonstrated the lowest inter-limb correlations with the majority at low and moderate values and a large range from .60 to .86.

3.2. Intra-limb coordination

Table 4 displays the mean and standard deviation for the intra-limb correlation coefficients for all joint comparisons (LA vs. LK, LK vs. LH, etc.) and all infants. Intra-limb correlation values were high for I4, but ranged from moderate to high (.72 to .94) for I2. Approximately half of the intra-limb correlations were high for I1 and the remainder were low to moderate correlations. I3 only demonstrated one instance of high correlations, while the majority were low to moderate values.

Left ankle-knee correlation was significantly higher for C1 than all other conditions (all comparisons $p \leq 0.003$) for I1 and I2; however, left ankle-knee correlation for I2 was also significantly higher in C2 than C3 ($p = 0.009$) and C4 was significantly higher than C3 ($p = 0.019$). Right ankle-knee correlations for I1 were significantly higher in C1 than C3 and C4 ($p \leq 0.006$). Right ankle-knee correlations were significantly different across all conditions for I2 ($p \leq 0.003$). I3 and I4 only had significant left ankle-knee correlation differences between C1 and C3, and C2 and C3 (both comparisons $p < 0.0005$). Right ankle-knee correlations were not significantly different across conditions for I3; however, I4 had significant differences between C2 and C3, and C3 and C4 (all comparisons $p \leq 0.013$).

For I1 and I2, left knee-hip correlations were significantly lower in C2 when compared to the other conditions (all comparisons $p \leq 0.002$), with additional significant differences between C3 and C4 ($p < 0.036$ for I1; $p < 0.001$ for I2). I1 had significant right knee-hip correlation differences between C1 and C2 ($p = 0.01$), while significant right knee-hip correlations were found across all conditions, except C2 and C3, for I2 (all comparisons $p < 0.0005$). For I3, left knee-hip correlation was significantly lower in C3 than C2 and C1 (both comparisons $p \leq 0.004$). No significant right knee-hip correlation differences were found for I3. Left knee-hip correlations for I4 were the highest in C1, where all condition comparisons, except C3 and C4, were significantly different (all comparisons $p \leq 0.045$). Right knee-hip correlations were significantly different across all condition comparisons against C2 for I4 (all $p < 0.0005$).

For I1, all left ankle-hip correlations were significantly different across conditions except between C1 and C4 (all comparisons $p < 0.0005$). Right ankle-hip correlations were significantly different between C1 and C3 ($p = 0.045$) C1 and C4 ($p = 0.004$), and C2 and C4 ($p = 0.004$) for I1. All condition comparisons except for the comparison between C1 and C2 were significantly different for I2 for the left ankle-hip correlations ($p \leq 0.027$). Right ankle-hip correlations were significantly different across all conditions for I2 ($p \leq 0.039$), except between C1 and C4. Left ankle-hip correlations were significantly different between C1 and C3 for I3 ($p = 0.028$), while right ankle-hip correlations were not significantly different across conditions ($p \geq 0.949$). Left ankle-hip correlations for I4 had significant differences between C1 and the other conditions ($p < 0.0005$), while right ankle-hip correlations were significantly different between all conditions except C2 and C4 ($p \leq 0.043$).

3.3. Phase planes

A trend of similar changes was found between I1, I2 and I4, therefore, their results were grouped together (Fig. 4) while the remaining infant is reported separately (Fig. 5). Infant 1, 2 and 4 began C1 with patterns that deviated moderately from spring-like circular phase planes and by C4 demonstrated increasingly spring-like circular phase planes. The majority (more than 50%) of phase plane mechanisms for I1, 2 and 4 in C1 for the ankle joints had phase plane shapes that deviated moderately from a circular pattern (dimples). Knee and hip joints in C1 were variable for all infants, but demonstrated phase planes with greater deviations from circular spring-like portraits. In C2, the predominant phase planes at the ankles for infants were of phase portraits that moderately to lightly deviated from spring-like patterns, dimples followed by convexes. Phase planes with larger deviations from spring-like patterns (trisectrices and dimples) for I1 and I2, and phase plane patterns that deviated less from a circular pattern for I4 dominated the phase planes at the knee joints. A minimum of 37.4% of trisectrices were found at the hip joints in C2 for all infants, while the rest of time the hip joints varied among the other shapes. For C3, all the infants began to display higher percentages of more spring-like phase plane shapes at the ankle, knee and hip joints. In C4, it was evident that the phase planes were spring-like, predominately convexes followed by circular patterns and spirals.

In comparison to the other infants who demonstrated a higher percentage of dimpled phase planes, I3 began with more spring-like phase planes and her behaviour remained spring-like throughout all the conditions. For C1 ankle, knee and hip joints displayed a range of 33.2% to 45.3% of convex phase planes (Fig. 5). For C2, the ankle joints became less spring-like, as there was higher percentage of dimpled phase plane patterns. The knee joints remained the same with an equal percentage of convexes with a slight increase in phase

planes that deviated from spring-like patterns (trisectrices) by 8% from C1. The hip joints increased in convex phase planes from the previous condition. Once I3 performed C3 the phase plane distribution of the ankle joints demonstrated an increase in convex phase planes to 52.6% (right) and 49.5% (left), however, there was also a slight increase trisectrices of 1–2% for each joint. The right knee joint displayed a reduction in convexes from C2 with 37.9%. Convexes, dimples and trisectrices, however, were equally distributed at the left knee joint. The hip joints had 32.6% of convex phase planes, followed by dimple phase planes. As a result, I3 did not demonstrate increasing spring-like behaviour as the natural frequency of the SS increased.

4. Discussion

The objective of this study was to investigate the mechanisms of control and the leg movements of four infants as they bounced to increasing natural frequencies. As hypothesized, I1, I2 and I4 decreased their lower limb joint ranges of motion as the natural frequency of the spring system increased. Similar findings of decreasing vertical T8 displacement were seen for these infants (Habib Perez et al., unpublished). In all conditions, I3 demonstrated relatively equal joint ranges of motion and equal bounce heights. In addition, I3 produced lower joint ranges than the other infants. A similar finding was seen for I3 in vertical T8 displacement when compared to the other infants (Habib Perez et al., unpublished). For each infant, ranges of motion were always greater on one side for each joint since the majority of left and right ranges were significantly different. Interestingly, the inter-limb correlation values were low/moderate to high for I3 (.60 to .86) and moderate to high for the remaining infants (.72 to .96), suggesting that movements were synchronous and was achieved with one side compensating by modifying the ROM. These results are similar

to participants who adapted oscillatory bimanual leg and arm movements of a timing task when masses were placed on a single limb (Jeka & Kelso, 1995). Bimanual oscillatory movement amplitudes of an arm and leg decreased when a mass was independently added to either limb; however, mean cycle amplitude of the leg also decreased when then mass was added to the arm. Since both the arm and leg decreased in amplitude, both limbs compensated the additional load to a single limb (Jeka & Kelso, 1995). The underlying adaptation that infants use, whether they regulate bounce frequency or time on the ground, is related to the number of joints that compensate to maintain synchrony.

There were no ipsilateral compensatory patterns at the ankle or hip joints for I1 and I2; yet, the left knee always traveled a greater ROM for the frequency changes for I1, while the right knee traveled a larger range for the frequency changes for I2. On the other hand, for I4, the right hip always traveled a larger ROM in all conditions, while the ankles and knees did not demonstrate any consistent patterns. It was evident that one joint, either on the left or right side, always compensated for the frequency changes of the system when infants matched the natural frequency of the spring system. In contrast, when infants regulated another timing component, such as the time on the ground, one leg was dominant. With greater ranges of motion on the right side, I3 displayed ipsilateral dominance in all joints. Therefore, we propose that the methods that infants adapt to the frequency changes of the spring system are contingent on the number of joints that always compensate for the frequency changes.

Other differences between the infants who adapted to the frequency changes using either strategy are the phase planes produced. Infants 1, 2 and 4 shifted their mechanics from less to more spring-like phase planes as the natural frequency of the system increased,

whereas I3 began more spring-like and produced relatively constant phase planes across conditions.

Infants 1, 2 and 4 began bouncing during C1 with a very small percentage of circular spring-like patterns, suggesting that their joints did not behave like springs in series. These infants also produced pronounced asymmetrical sinusoidal joint ranges of motion, as seen in Figure 2A at the ankle, knee and hip for I4. Infant 3, however, began with a higher percentage of circular patterns and smaller ranges of motion (Fig. 3A and 5). Visually, the joint ranges of motion that I3 produced in C1 contained smaller amplitudes, but displayed more symmetry (Fig. 3A). The limaçon position in the phase plane shapes occurred when all infants were in contact with the ground, which was at the same time that all joint ranges reach their minima. As the ranges of motion decreased across conditions for I1, I2 and I4, the minimum of each joint's ROM was less pronounced. The same infants also produced more circular spring-like patterns as spring frequency increased, suggesting that the changes in phase planes with less spring-like phase portrait (dimples, cardioids and trisectrices) to phase planes with little or no deviation of a circular spring-like patterns (convexes, spirals and circles) was partially attributed to the symmetry of the joint ranges of motion. This is confirmed with the findings for I3 whose joint ranges displayed more symmetry than the other infants and remained equal across conditions. Additionally, there were no phase plane pattern shifts from limaçon shapes to circular shapes as the frequency of the spring system increased for I3.

Though Vallis (1998) suggested that different phase plane patterns corresponded to the load cell loading patterns of infant bouncers, our study also demonstrated that changes in phase plane patterns in infant bouncers resulted from the joint ROM when the infants were in contact with the ground. This indicates that two factors contributed to the changes in phase

planes from a circle to a trisectrix limaçon, decreasing to increasing ROM for movement reversal to occur (Winstein & Garfinkle, 1989) and the amount of force applied (amplitude of the VGRFs). Vallis reported that the underlying control mechanisms seen within phase planes in infants with asymmetrical (slanted) load cell loading were circular phase planes, while infants with symmetrical sinusoidal loading patterns produced limaçons. With the findings of our previous study (Habib Perez et al. unpublished), the current study demonstrated that increasing VGRFs and the symmetry of joint ROM affect the patterns of the phase planes.

In addition, Vallis (1998) showed that phase planes were not dependent on skill level, all infants produced circular and limaçon patterns; however, less skilled infants were very variable. It was suggested that moderately skilled and skilled infants were in different attractor states as they occupied a preferred space in the plot when the infants produced consistent and overlapping phase plane shapes. While the current study does not analyze plot space, our results suggest through frequency counts that I1, I2 and I4 were in two stable attractor states in the first and last conditions. The prominent dimpled limaçon phase planes for C1 and convex and circular phase planes for C4 confirm that these infants were in stable attractors. Circular phase planes, however, have been shown to be less stable (Vallis, 1998) and can be supported with the higher bounce frequency variability found in C4 (Habib Perez et al. unpublished). The variable distribution of the different phase planes seen in C2 and C3 for these infants suggests that infants were not in stable attractors. A similar variability of phase planes was evident in I3 across all conditions, which indicates that I3 was never in a stable attractor, but did demonstrate a preferred mechanism of control.

Our findings contradict the original prediction of Goldfield, Kay and Warren (1993) who stated that infants who matched the resonant frequency of the system had legs that

behaved like springs. Using Goldfield et al. (1993) reasoning, circular phase planes would not be the predominant pattern found for I3 since she produced bounce frequencies that did not match the natural frequency of the system. However, the circular phase plane patterns across conditions displayed by I3 demonstrated that her legs behaved like springs. I3's legs behaved partly spring-like to entirely spring-like, as the majority of phase planes represented phase portraits of convex limaçons, circles and spirals. Although infants are not always in a stable attractor state, their legs behave spring-like regardless of how they adapt to the frequency changes of the system. The method in which I3 adapts to the frequency changes of the system suggests that there may be a difference in skill level.

Another parameter that suggests that I3 displays different bouncing skill levels in comparison to the other infants are the intra-limb correlations. Low to moderate intra-limb correlations were seen for the majority of intra-limb couplings of I3. In addition, I3 demonstrated the highest values of intra-limb variability, suggesting that her leg coordination was not as stable as the other infants. Whereas, I1, I2 and I4 demonstrated only nine, six and one incidences, respectively, where intra-limb correlations were moderate. The majority of these incidences for the three infants occurred when an angular position was paired with the hip joint. We suspect that this may be caused by the limitation the harness imposes on the infant's hip ROM, as the infant remained in a sitting position with slight hip flexion throughout their bouncing. Contrary to Jensen et al. (1994), who reported a task constraint (postural) affect on intra-limb synchrony in kicking, the current study did not observe a task constraint (frequency change) affect on the infants. Placing infants in a vertical position when bouncing, however, resulted in one joint always compensating by traveling a greater ROM.

In our previous study (Habib Perez et al., unpublished), we reported that infants adapted to the frequency changes of the spring system using one of two strategies. The current study demonstrated that underlying mechanisms of control pertain to each bouncing strategy. Across conditions, I1, I2 and I4 reduced all joint ranges of motion to achieve increased bounce frequency and match the increasing natural frequency of each system. As a result, phase plane patterns were altered when joint ROM and VGRFs decreased, and were not a result of the load cell patterns as once suggested (Vallis, 1998). Infant's 1, 2 and 4 began bouncing in one stable attractor state and ended in a less stable attractor state. This was confirmed by the increasing bounce frequency variability from C1 to C4 (Habib Perez et al., unpublished), when high percentages of non-circular limaçons to spring-like phase-phase patterns were seen across conditions. For infants who used strategy one, we propose that the shift from non spring-like to circular spring-like patterns can perhaps also be attributed to the reduction in VGRFs, as this deviation occurred when the infant was in contact with the ground. To determine whether the amplitude of VGRFs contributed to the deformation of a circular phase plane, quantitative polar statistics analyses of phase planes are required.

Moderate to high inter-limb correlation values for I1, I2 and I4 indicate that infants moved their legs in synchrony, which attributed to the infants' ability to regulate bounce frequency. Low to moderate inter-limb correlations, in the case of I3, limited the ability to regulate the ability to regulate bounce frequency; however, I3 was still able to regulate the time on the ground. Since I3 was able to adapt to the frequency changes of the system, the strategy to adapt to the frequency changes indicated a different bouncing skill level. The combination of lower inter- and intra-limb correlations and smaller joint ranges of motion may quantitatively identify skill level.

5. Conclusion

The infants responded to the frequency modifications of the spring system using one of two timing strategies. Infants 1,2 and 4 regulated bounce frequency and bounced near the natural frequency of system produced well coordinated inter- and intra-limb leg movements. As a result of infants matching their bounce frequency to the natural frequency of the system, joint ranges of motion were reduced across conditions and increased in symmetry, which affected the underlying mechanics of each joint as seen in the phase planes.

In contrast, I3 appeared to regulate the time on the ground and was less coordinated. She did not reduce her joint ranges of motion or symmetry. In turn, underlying mechanics in phase planes do not change as a function of the frequency changes of the system. Although our findings provide insight on how typically developing infants adapt to environmental constraints, further research is required with a larger population to corroborate whether strategy of adaptation is related to skill level. Furthermore, quantitative phase plane analysis would advance our understanding of whether the applications of VGRFs contribute to a divergence of a spring-like circular phase plane.

Table 1 - Phase plane shapes and coding properties







Phase plane shapes	Phase plane properties
	<p>A circle was defined as circular without flat edges. This represents no deviation away from spring system.</p>
	<p>If circular amplitude decreased it carried the properties of a spiral. It was considered to be a spiral when decreasing circular overlap was present.</p>
	<p>A convex limaçon with a flat edge, and the flat edge cannot contain a dimple. If it has a flat edge but a dimple on the outside of the circular area then it is considered convex.</p>
	<p>A dimpled limaçon must be within the circular area.</p>
	<p>A cardioid limaçon with a cusp. Anything that was not acute was considered a dimple. The cusp needed to be centrally located within the graph.</p>
	<p>A trisectrix limaçon having a loop. If plot began with a loop, then it was coded as a trisectrix</p>
<p>Other</p>	<p>Any shape that does not fit into the other 6 codes</p>

Table 2 - Inter-limb correlation coefficients for the right and left ankle, knee and hip joints for all infants.

Infant	Condition	Inter-limb Correlations		
		Ankle	Knee	Hip
I1	1	0.92 ±.15	0.90 ±.15	0.91 ±.14
	2	0.90 ±.16	0.75 ±.36	0.79 ±.25
	3	0.72 ±.49	0.73 ±.41	0.85 ±.26
	4	0.76 ±.41	0.75 ±.42	0.79 ±.40
I2	1	0.96 ±.05	0.90 ±.11	0.94 ±.05
	2	0.94 ±.11	0.89 ±.19	0.90 ±.15
	3	0.92 ±.15	0.91 ±.15	0.92 ±.11
	4	0.92 ±.12	0.94 ±.11	0.95 ±.08
I3	1	0.76 ±.32	0.78 ±.29	0.86 ±.22
	2	0.72 ±.40	0.84 ±.26	0.81 ±.29
	3	0.66 ±.42	0.60 ±.42	0.77 ±.30
I4	1	0.94 ±.12	0.92 ±.18	0.93 ±.14
	2	0.94 ±.14	0.96 ±.05	0.91 ±.13
	3	0.88 ±.23	0.89 ±.19	0.81 ±.22
	4	0.93 ±.13	0.93 ±.12	0.85 ±.19

Note: Bolded text – strong/high correlations (>0.8); regular text – moderate correlation (0.7-0.8); grey text – low correlations (0.5-0.7).

Table 3 - *p*-values of pair-wise comparisons for ranges of motion and inter-limb correlations for all infants. (L/R = left/right, A/K/H = ankle, knee, hip)

		Infant 1							
		Range of Motion							
Pair-wise comparisons	LA	RA		LK	RK		LH	RH	
C1 – C2	<0.0005	<0.0005		<0.0005	<0.0005	*	<0.0005	<0.0005	**
C1 – C3	<0.0005	<0.0005	***	<0.0005	<0.0005	*	<0.0005	<0.0005	
C1 – C4	<0.0005	<0.0005	***	<0.0005	<0.0005	*	<0.0005	<0.0005	**
C2 – C3	<0.0005	<0.0005	***	0.173	1.000		0.014	0.143	
C2 – C4	<0.0005	<0.0005		<0.0005	<0.0005		<0.0005	<0.0005	
C3 – C4	1.000	0.001		<0.0005	<0.0005		0.074	<0.0005	
		Infant 2							
		Range of Motion							
Pair-wise comparisons	LA	RA		LK	RK		LH	RH	
C1 – C2	<0.0005	<0.0005	**	1.000	<0.0005		<0.0005	<0.0005	**
C1 – C3	<0.0005	<0.0005	**	0.444	<0.0005		0.158	<0.0005	**
C1 – C4	<0.0005	<0.0005	**	<0.0005	<0.0005	*	<0.0005	<0.0005	
C2 – C3	<0.0005	0.559		0.054	<0.0005		<0.0005	1.000	
C2 – C4	<0.0005	<0.0005		<0.0005	<0.0005	*	<0.0005	<0.0005	**
C3 – C4	<0.0005	<0.0005		<0.0005	<0.0005	***	<0.0005	<0.0005	**
		Infant 3							
		Range of Motion							
Pair-wise comparisons	LA	RA		LK	RK		LH	RH	
C1 – C2	0.001	0.027		0.844	0.172		0.069	1.000	
C1 – C3	1.000	0.814		<0.0005	<0.0005	**	0.002	0.001	
C2 – C3	0.002	0.295		<0.0005	<0.0005	**	<0.0005	0.001	
		Infant 4							
		Range of Motion							
Pair-wise comparisons	LA	RA		LK	RK		LH	RH	
C1 – C2	<0.0005	<0.0005		0.048	1.000	*	<0.0005	<0.0005	
C1 – C3	<0.0005	<0.0005	*	<0.0005	<0.0005		<0.0005	<0.0005	*
C1 – C4	<0.0005	<0.0005	*	<0.0005	<0.0005		<0.0005	<0.0005	*
C2 – C3	<0.0005	<0.0005	*	<0.0005	<0.0005	*	<0.0005	<0.0005	*
C2 – C4	<0.0005	<0.0005	*	<0.0005	<0.0005	*	<0.0005	<0.0005	*
C3 – C4	<0.0005	<0.0005	*	<0.0005	<0.0005	*	<0.0005	<0.0005	*

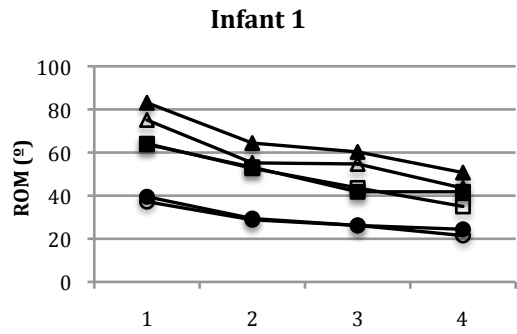
Note: Asterisks indicate significant differences in inter-limb correlations for the ankle, knee and hip, * (p<0.001), ** (p<0.010), and *** (p<0.05).

Table 4 – Intra-limb correlation coefficients for the left and right leg for all infants. (L/R = left/right, A/K/H = ankle, knee, hip)

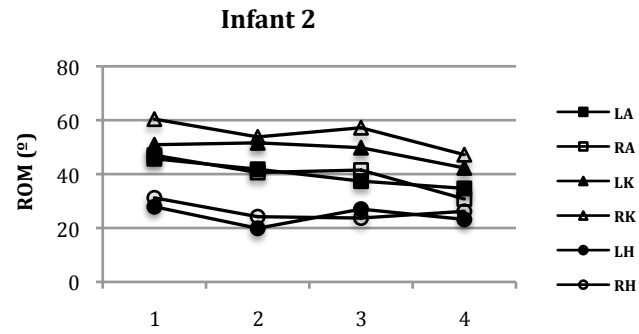
Infant	Condition	Intra-limb Correlations					
		LA-LK	LK-LH	LA-LH	RA-RK	RK-RH	RA-RH
I1	1	0.93 ±.07	0.85 ±.20	0.83 ±.21	0.87 ±.16	0.84 ±.19	0.87 ±.14
	2	0.81 ±.29	0.68 ±.34	0.67 ±.29	0.83 ±.23	0.74 ±.35	0.81 ±.20
	3	0.80 ±.31	0.81 ±.27	0.76 ±.32	0.76 ±.33	0.77 ±.35	0.75 ±.36
	4	0.85 ±.23	0.89 ±.18	0.86 ±.19	0.69 ±.44	0.75 ±.38	0.61 ±.49
I2	1	0.80 ±.17	0.88 ±.12	0.85 ±.12	0.83 ±.12	0.86 ±.15	0.90 ±.11
	2	0.90 ±.14	0.79 ±.22	0.80 ±.19	0.88 ±.18	0.79 ±.25	0.77 ±.26
	3	0.86 ±.17	0.84 ±.21	0.74 ±.26	0.92 ±.09	0.76 ±.24	0.72 ±.23
	4	0.90 ±.14	0.91 ±.17	0.87 ±.15	0.94 ±.09	0.93 ±.14	0.91 ±.13
I3	1	0.75 ±.25	0.82 ±.22	0.75 ±.27	0.73 ±.28	0.78 ±.25	0.75 ±.31
	2	0.73 ±.28	0.74 ±.32	0.62 ±.47	0.76 ±.29	0.71 ±.34	0.72 ±.38
	3	0.41 ±.50	0.56 ±.46	0.61 ±.41	0.69 ±.38	0.78 ±.32	0.71 ±.31
I4	1	0.92 ±.17	0.93 ±.14	0.89 ±.15	0.94 ±.15	0.92 ±.13	0.93 ±.09
	2	0.92 ±.12	0.86 ±.16	0.81 ±.16	0.95 ±.08	0.95 ±.06	0.90 ±.08
	3	0.86 ±.22	0.82 ±.25	0.77 ±.26	0.92 ±.10	0.89 ±.14	0.87 ±.13
	4	0.89 ±.17	0.81 ±.27	0.82 ±.22	0.94 ±.09	0.92 ±.13	0.90 ±.13

Note: Bolded text – strong/high correlations (>0.8); regular text – moderate correlation (0.7-0.8); grey text – low correlations (0.5-0.7). LA, LK and LH denotes left ankle, left knee and left hip, while RA, RK and RH denotes right ankle, right knee and right hip.

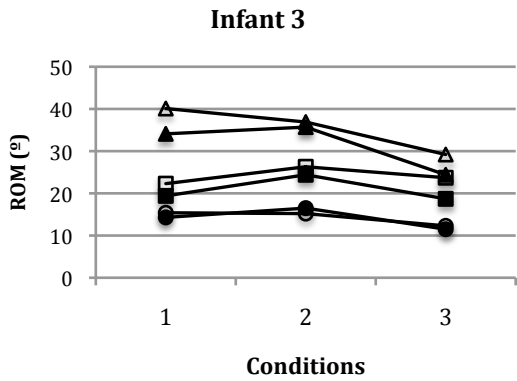
A



B



C



D

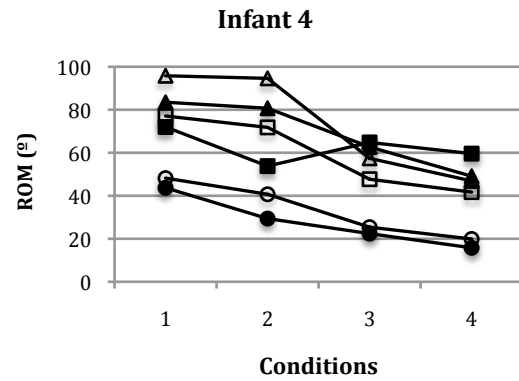


Figure 1 – Mean range of motion (ROM) for the left and right ankles, knees and hips for Infant 1 (A), 2 (B), 3 (C) and 4 (D) across conditions.

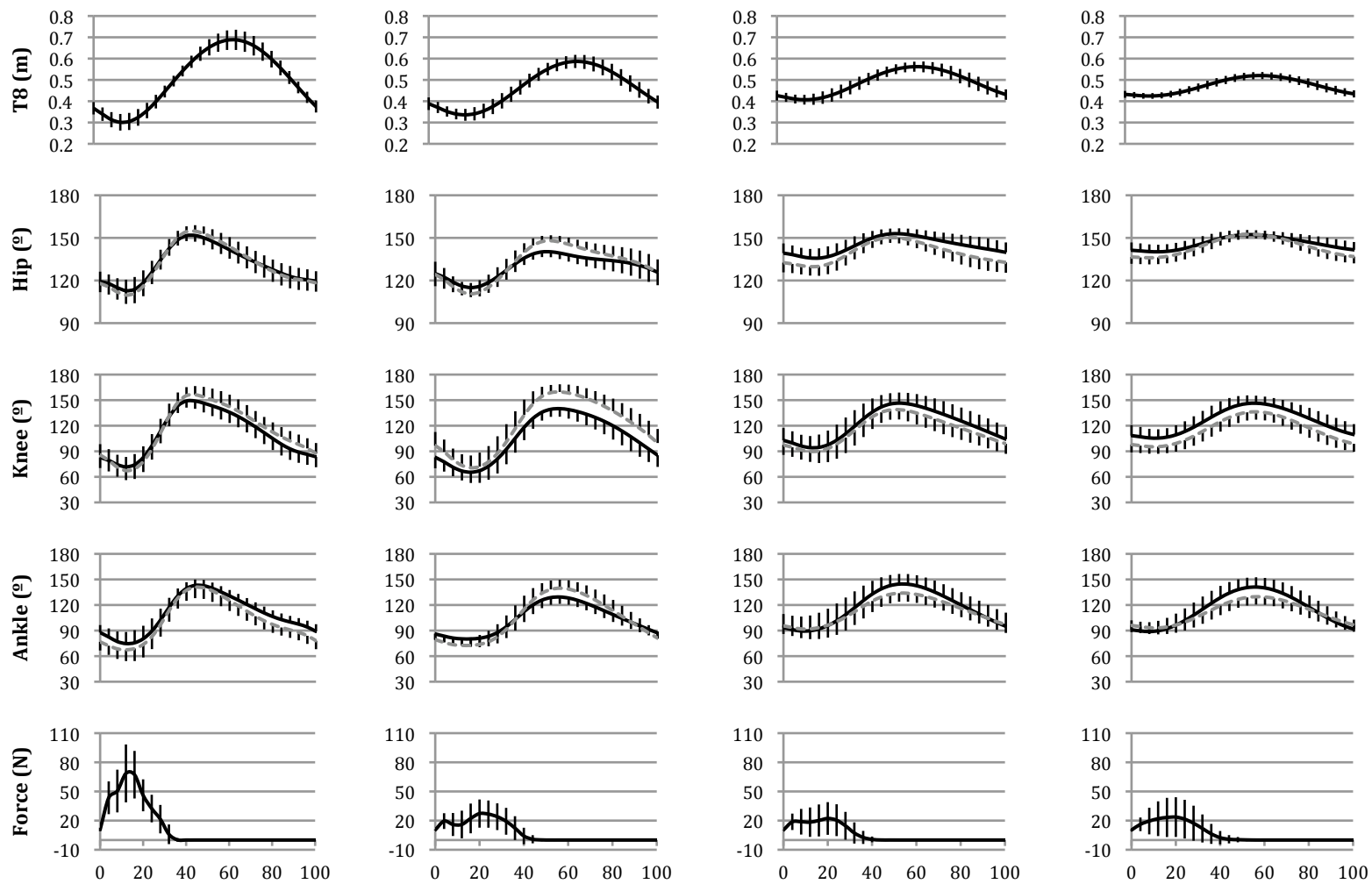


Figure 2 A

Figure 2 B

Figure 2 C

Figure 2 D

Figure 2A-D – Means and SD for the ensemble-averaged and time normalized T8 vertical displacement, ankle, knee and hip ranges of motion and vertical ground reaction forces for 14 in all four conditions. 2A – Condition 1; 2B – Condition 2; 2C – Condition 3; 2D – Condition 4. Dashed lines in joint ROM denote right leg, while solid black line represents left leg.

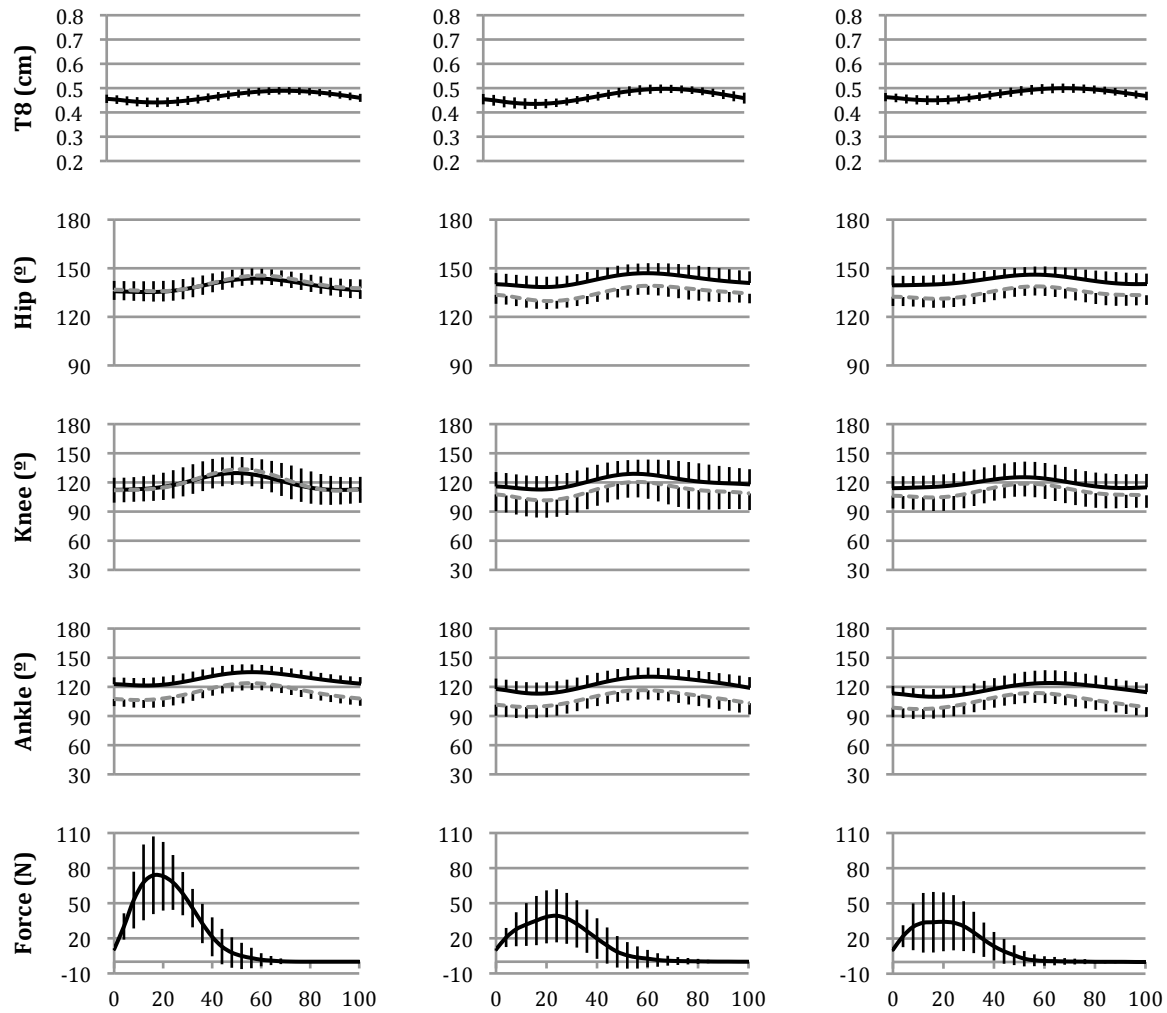


Figure 3 A

Figure 3 B

Figure 3 C

Figure 3A-C Means and SD for the ensemble-averaged and time normalized T8 vertical displacement, ankle, knee and hip ranges of motion and vertical ground reaction forces for I3 in all conditions. 3A – Condition 1; 3B – Condition 2; 3C – Condition 3. Dashed lines in joint ROM denote right leg, while solid black line represents left leg.

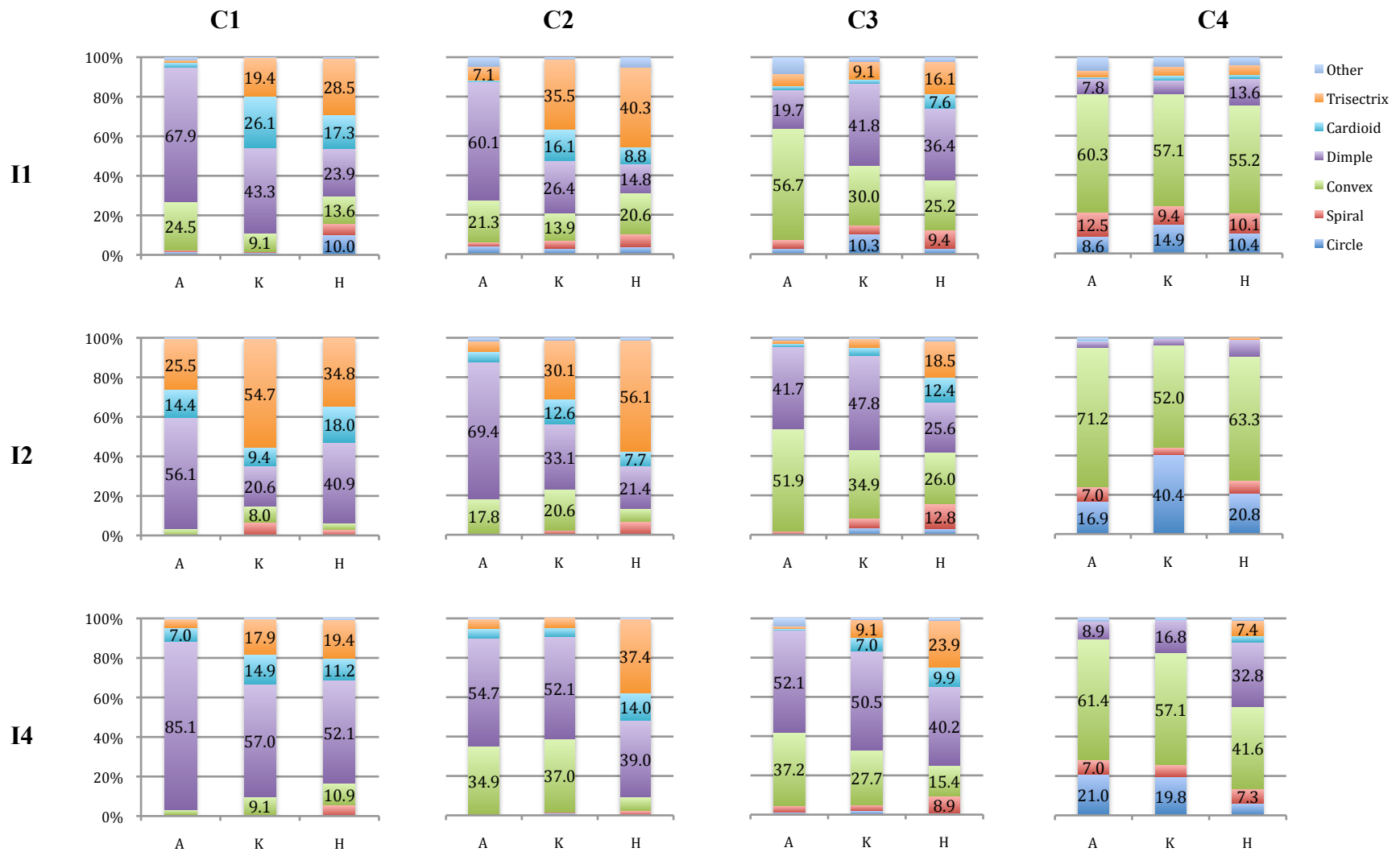


Figure 4 – Phase plane distribution for I1, I2 and I4; all infants using strategy 1.

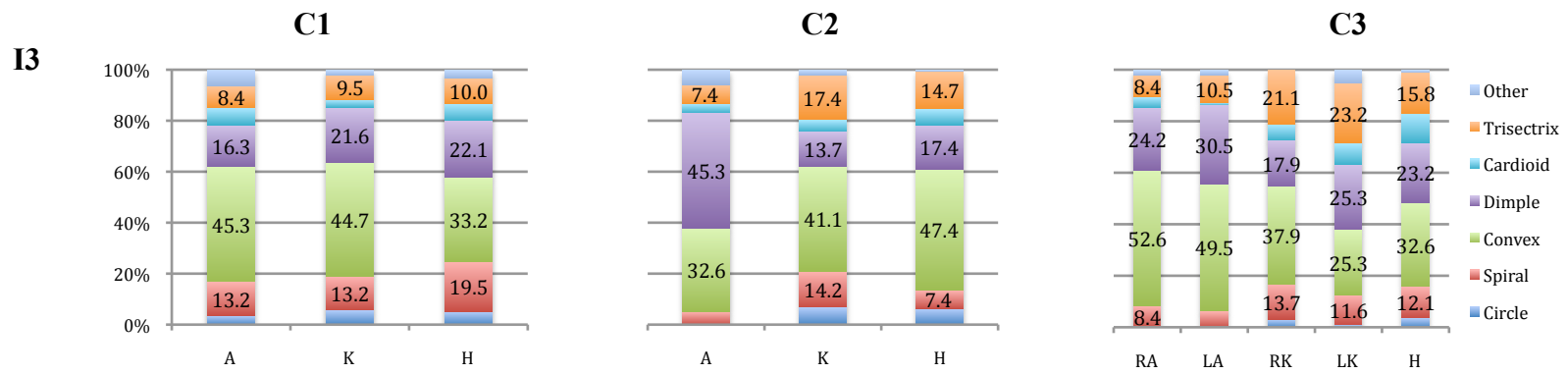


Figure 5 – Phase plane distribution for I3, using strategy 2.

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CHAPTER SIX

GENERAL DISCUSSION

The study of infant bouncing at different spring frequencies yielded interesting findings of how infants adapt to different environmental constraints. Typically developing pre-walking infants were chosen for this study because the study focused on understanding how infant bouncing changed in different environmental conditions. Unlike previous research in infant bouncing (Foo, Goldfield, Kay, & Warren, 2001; Goldfield, Kay, & Warren, 1993; Vallis, 1998), the current study used four spring conditions of increasing natural spring frequency over two testing sessions and enabled the investigation of adaptive movement behaviour without any developmental maturation. A dynamical system perspective was used to understand how infants adapt to spring frequency constraints.

The current study can reject the primary null hypothesis, which was that bounce frequency would not be significantly different across spring frequencies, as bounce frequency was significantly different across spring frequencies in all infants. We can also reject all secondary null hypotheses, which outlined that the duration of each bounce, amplitude of each bounce, peak vertical ground reaction forces, ankle, knee, hip joint ranges of motion, and inter- and intra-limb coordination would not be significantly different across spring frequencies. For all the aforementioned variables, the current study found significant differences across conditions. Through analyses of kinematic and kinetic variables, the current study demonstrated multiple elements that outline an infant's ability to adapt to natural frequency changes of the spring system. The

manuscripts used a theoretical framework—dynamical systems theory—and discussed how environmental constraints affected infant bouncing, a closed-looped task-specific behaviour. The findings were viewed in terms of providing a foundation of future research; however, additional suggestions are discussed here.

Article Findings

In the first article, we contributed novel findings to the understanding of infant bouncing. Infants bounced at a higher bounce frequency than the natural frequency of the spring system, which indicates that infants do not solely behave like a mass-spring system. Regardless of how infants adapt to the changes of the spring system, all infants' VGRFs decreased as the natural frequency of the system increased. Based on the asymmetrical loading patterns recorded, we suggest that infants were regulating a timing component similar to when musicians were instructed to oscillate their fingers at given increasing frequencies (Balasubramaniam, Wing, & Daffertshofer, 2004). From the findings of Balasubramaniam and colleagues (2004) and the applications to the current study, it is believed that asymmetrical loading occurs as error correction to maintain synchrony. In bouncing, when infants initially touch the ground they load the ground faster and unload slower to correct for the asymmetry but maintain synchrony. While infants bounce and adapt to the frequency changes of the system they regulate a timing component using one of two strategies. The first strategy found infants regulating bounce frequency with an additional constant of approximately 0.2 Hz above the natural frequency of the system. To match the natural frequency of the system, infants reduced their vertical trunk displacement. The second bouncing strategy seen in a single infant,

found that the amount of time on the ground was regulated and lead to equal trunk displacement across conditions.

Further analyses of lower extremity kinematics were conducted to understand the mechanisms of control and leg movements of infant bouncers across increasing spring frequencies. Using common measures of previous literature (Balasubramaniam et al., 2004; Fetters, Chen, Jonsdottir, & Tronick, 2004; Heriza, 1988; 1991; Piek & Gasson, 1999; Vallis, 1998; Winstein & Garfinkle, 1989) joint range of motion (ROM), intra- and inter-limb coordination were analyzed. Joint ranges of motion decreased like the vertical trunk displacement of infants who regulated bounce frequency across conditions. For these infants to regulate bounce frequency inter-limb coordination was required to be of moderate to high synchrony. In the infant who regulated time on the ground rather than bounce frequency, inter-limb coordination was not as synchronous. A novel finding of the second article was that the joints of the lower extremity always compensated for the frequency changes of the spring system, as one side always traveled a greater ROM than the other. These findings were similar to those of Jeka and Kelso (1995) when they reported that both the arm and leg compensated for additional weight placed on a single limb during bimanual oscillatory movements increasing frequencies. Through joint ROM and inter-limb coordination we were able to identify a difference in skill level between the two strategies that infants used to adapt to the frequency changes of the system. Unlike Vallis (1998), who initially qualitatively categorized infant bouncing skill levels prior to analyzing any kinematic variables, the current study was able to provide more tangible values and identify skill level. Nonetheless, our findings were similar in that moderate skilled bouncers produce poorer inter-limb coordination to that of skilled infant

bouncers; however, Vallis' (1998) analysis was only qualitatively driven. As a result of skill level, intra-limb coordination differed between infants who used strategy one and strategy two. Finally, through use of novel qualitative phase plane analysis, we identified that all infants produce all types of phase plane patterns. Infants who adjust to frequency perturbations by matching the natural frequency of the spring system, however, begin and end in a stable attractor state in conditions one and four, while infants who regulate time on the ground remain in relatively stable attractor state.

FUTURE RECOMMENDATIONS

The results from the current study on infant bouncing have provided a foundation for future research, particularly by manipulating the environmental constraints. Since the current study only used four discrete spring conditions we do not know how infants transition from one frequency to another. To explore this gap, continuous frequencies changes are required with a spring or apparatus that allows such changes to occur.

A second suggestion for future research is to manipulate the sensory feedback information the infant receives when coming in contact with the ground. Infant bouncing is a closed-looped task-specific behaviour and requires online sensory feedback that prepares the infant's subsequent bounce. Altering the material of the ground using materials of various densities, such as placing a gel cushion versus dense foam, may help identify how and which materials inhibit infants in maintaining timing accuracy. Furthermore, placing different VGRF thresholds with a stimulus, such as a light or music, similar to the infants who learned to kick at specific angular thresholds (Angulo-Kinzler, Ulrich, & Thelen, 2002), would identify whether infants can adapt to different VGRF

task demands and identify at which amount does the VGRF affect a change in phase planes.

Moreover, it would be interesting to identify whether infant bouncing is driven by vestibular stimuli. Attaching a triaxial accelerometer to the infants' head with the use of a headband would provide information of the head's linear and angular acceleration in three directions (axes). The acceleration of head could be related to linear and angular acceleration of the vestibular system, and may explain what infants are intending to regulate and whether they are maintaining a constant acceleration.

CONTRIBUTIONS

Olinda Habib Perez developed, designed, and undertook this thesis, its analysis and writing. Dr. Heidi Sveistrup and Dr. Gordon Robertson supported all aspects of the study development, analysis and writing and provided advice and input into reviewing the final product. Dr. Natalie Baddour also supported all aspects of the study and in addition to reviewing the final product she contributed engineering expertise to the data analysis.

Coren Walters-Stewart, BEng, BSc, MSc 1) wrote all MATLAB programs 2) identified the shapes for qualitative analysis of phase planes based on polar coordinate equations and 3) coded all phase planes by visual inspection. Alexander Mineault-Guitard aided with data processing.

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Winstein, C. J., & Garfinkle, A. (1989). Qualitative Dynamics of Disordered Human
Locomotion: A preliminary investigation. *Journal of Motor Behavior, 21*(4), 373-
391.

APPENDICES

Appendix A

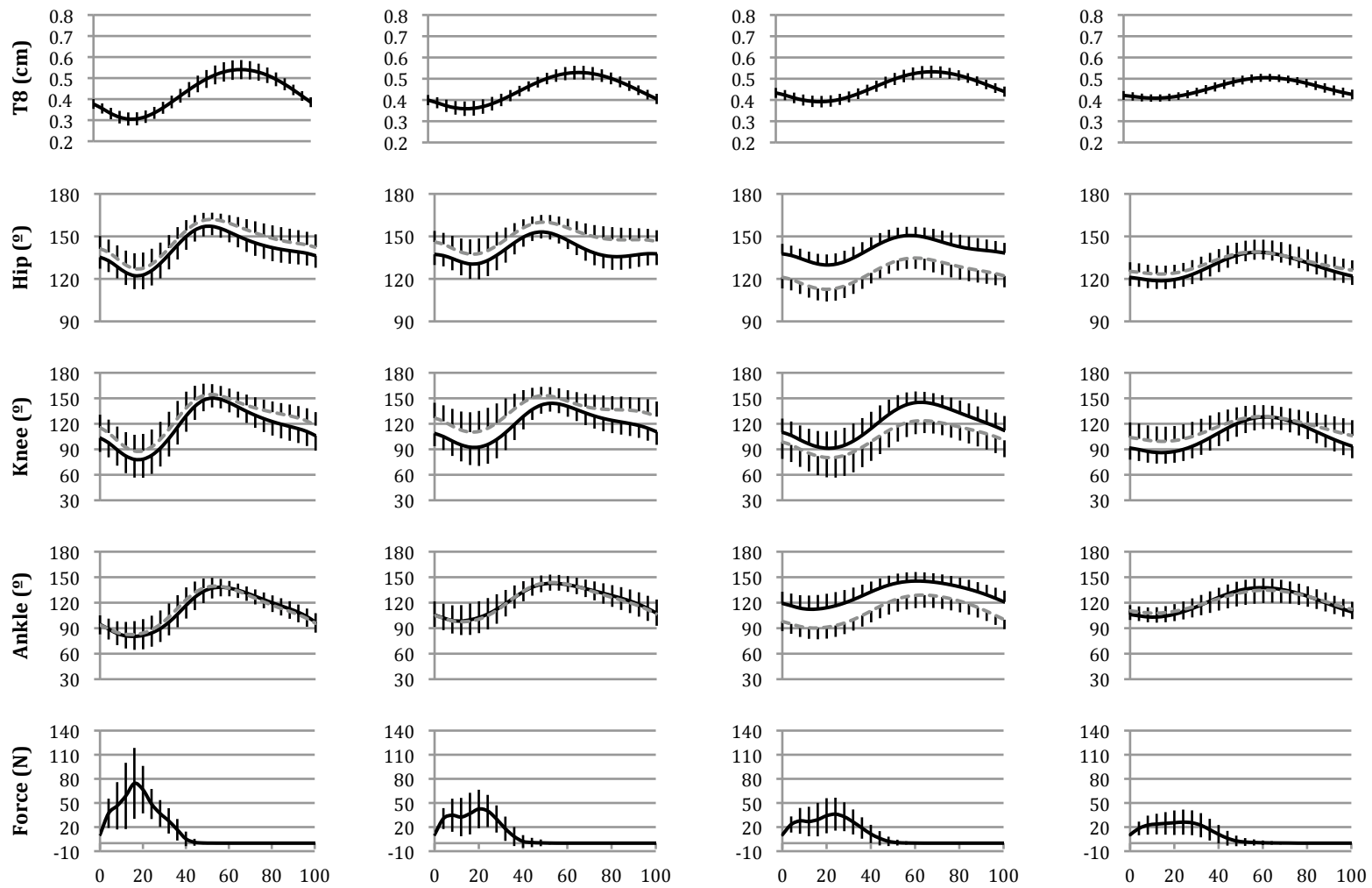


Figure 1 A

Figure 1 B

Figure 1 C

Figure 1 D

Figure 1A-D – Means and SD for the ensemble-averaged and time normalized T8 vertical displacement, ankle, knee and hip ranges of motion and vertical ground reaction forces for I1 in all four conditions. 1A – Condition 1; 1B – Condition 2; 1C – Condition 3; 1D – Condition 4. Dashed lines in joint ROM denote right leg, while solid black line represents left leg.

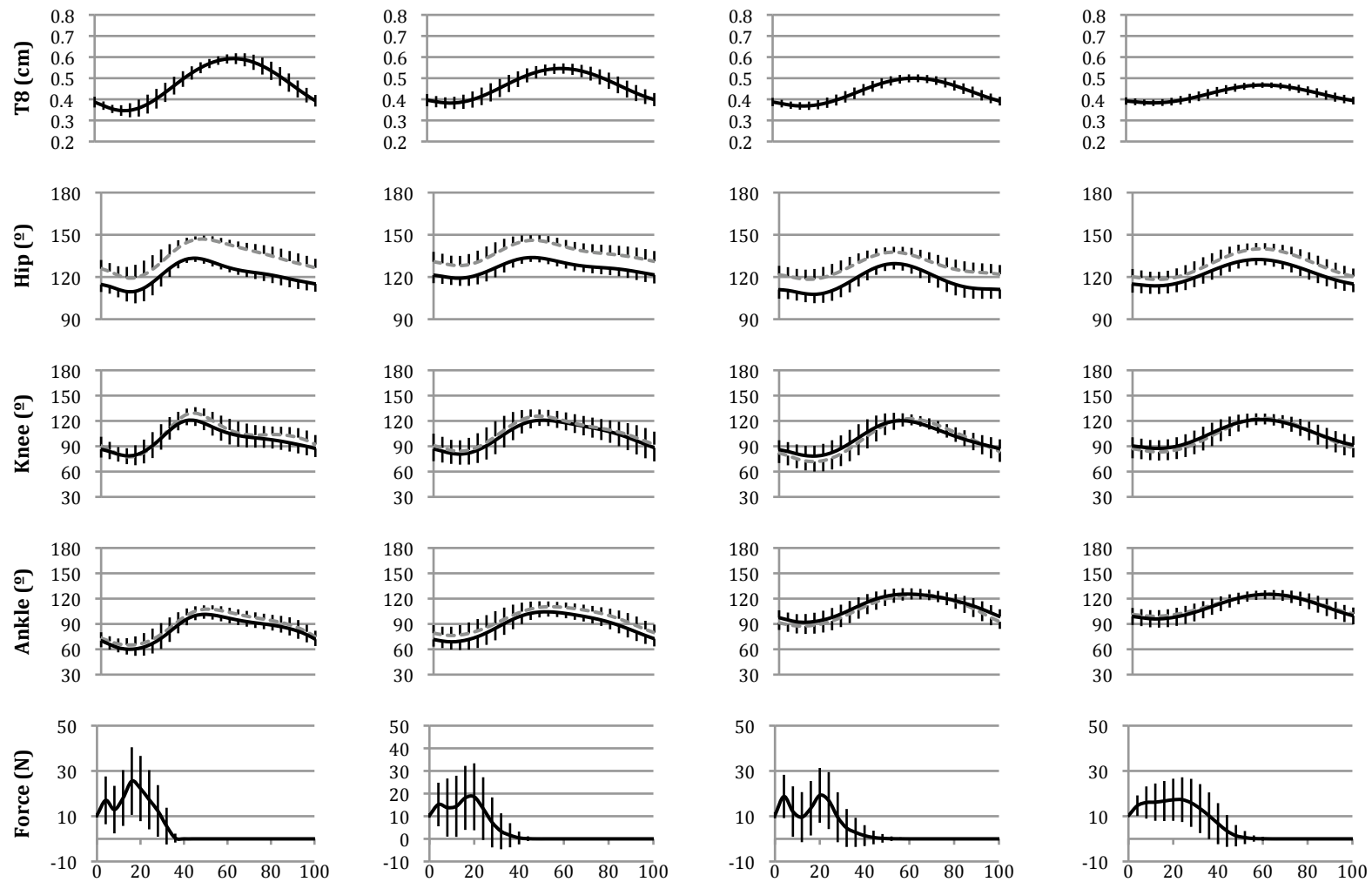


Figure 2A-D – Means and SD for the ensemble-averaged and time normalized T8 vertical displacement, ankle, knee and hip ranges of motion and vertical ground reaction forces for 12 in all four conditions. 2A – Condition 1; 2B – Condition 2; 2C – Condition 3; 2D – Condition 4. Dashed lines in joint ROM denote right leg, while solid black line represents left leg.

Appendix B

- File *importfile2.m* imports filtered vertical ground reaction data and kinematic data for Appendix A, B and C.

- File *processbounceforce.m* MATLAB script extracts the variables listed at the top beginning of the file (A-Q).

```
function importfile2(fileToRead1)
%IMPORTFILE2(FILETOREAD1)
% Imports data from the specified file
% FILETOREAD1: file to read

% Auto-generated by MATLAB on 07-May-2010 17:18:43

DELIMITER = '\t';
HEADERLINES = 5;

% Import the file
newData1 = importdata(fileToRead1, DELIMITER, HEADERLINES);

% Create new variables in the base workspace from those fields.
vars = fieldnames(newData1);
for i = 1:length(vars)
    assignin('base', vars{i}, newData1.(vars{i}));
end

-----

%Outputs the following variables to a file with a label corresponding to
%the source file, but with the suffix of F.csv

%Variables:
    %fulloutput=[A force start time|B force end time|C length of ground interval|D
length
    %of air interval|E total bounce period|F peak force time|G peak
force
    %force|H bounce frequency|I percent on ground|J percent in air|K percent when peak
    %occurs|L time into bounce cycle when peak force occurs|
    %M slope touchdown to peakforce |N slope peak force to toe off |
    %O time from peak force to toe off|P peak force normalized to baby weight|
    %Q peak force normalized to system weight]

close all
clear all

%z is the number of condition trial pairs (number of rows in the variables CT and trial)
for z=1:11

    %CT represents the condition (1 through 4), trial number, and number of
    %bouts per specified trial
    CT=[1 2 8;1 3 9;2 5 11;2 6 5;2 7 1;2 8 2;3 2 1;3 3 13;3 4 9;4 5 14;4 6 5];

    %the rows in the variable trial represent the bouts in the trial specified
    %by the condition and trial number in the same row of the matrix CT
    trial=[3 4 6 8 9 10 11 12 0 0 0 0 0 0;
1 2 3 4 5 6 7 8 9 0 0 0 0 0;
1 2 4 5 6 7 8 9 10 11 12 0 0 0;
1 2 3 5 6 0 0 0 0 0 0 0 0 0;
1 0 0 0 0 0 0 0 0 0 0 0 0 0;
1 2 0 0 0 0 0 0 0 0 0 0 0 0;
1 0 0 0 0 0 0 0 0 0 0 0 0 0;
1 2 4 5 7 8 9 11 12 14 16 18 19 0;
2 3 4 5 6 7 8 9 11 0 0 0 0 0;
1 2 3 4 5 6 7 8 9 10 11 12 13 14;
1 2 3 4 5 0 0 0 0 0 0 0 0 0];

    q=CT(z,1);%condition number
    r=CT(z,2);%trial number
```

```

for y=1:CT(z,3) %number of bouts
    %calls the appropriate file of raw data for the condition, trial and
    %bout specified
    s=trial(z,y);
    importfile2(strcat('IS05C',num2str(q),'T',num2str(r),num2str(s),'F.txt'))

    %clears previous data set
    tab1=[0; 0;];
    tab2=[0; 0;];
    gdinterval=[0; 0;];
    airinterval=[0; 0;];
    totalperiod=[0; 0;];
    pkforceindex=[0; 0;];
    pkforcevalue=[0; 0;];

    forcedata=[data(:,1) data(:,4)];%index number and total force
    frequency=600; %in Hz
    timestep=1/frequency;%in seconds

    m=1;
    o=1;

    %to determine the indices in which the force changes from 0 to a number(tab1) or
    %a number to zero (tab2)

    for n=2:length(forcedata)-1;
        if forcedata(n,2)>0
            if forcedata(n-1,2)==0
                %initialize tab1
                tab1(m)=forcedata(n,1);
                m=m+1;
            elseif forcedata(n+1,2)==0
                %initialize tab2
                tab2(o)=forcedata(n,1);
                o=o+1;
            else
                countnothing=1;
            end
        else
            countnothing=2;
        end
    end

    %removes the first full or partial bounce from the data set
    if tab2(1)>=tab1(1)
        tab2=tab2(2:length(tab2));
        tab1=tab1(2:length(tab1));
    else
        tab2=tab2(2:length(tab2));
        tab1=tab1(:);
    end

    if length(tab1)==length(tab2)
        tab1=[tab1(:);0];
    else
        tab1=tab1(:);
    end

    %determines the time intervals during which the child is on the
    %ground and during which the child is in the air(by index number)
    for b=1:(length(tab2))
        gdinterval(b,1)=tab2(b,1)-tab1(b,1);%ground interval
        airinterval(b,1)=tab1(b+1,1)-tab2(b,1);%air interval
        totalperiod(b,1)=gdinterval(b,1)+airinterval(b,1);%total interval of the
bounce cycle
    end

    %finds the peak force in any given bounce cycle, and the index at
    %which the peak force occurs

```

```

for c=1:length(tab2)
    d=forcedata(tab1(c):tab2(c),2);
    pkforce=max(d);
    for f=tab1(c):tab2(c)
        if forcedata(f,2)==pkforce
            pkindex=f;
        else
            countnothing=0;
        end
    end
    pkforceindex(c)=pkindex;
    pkforcevalue(c)=max(d);
end

%this section contains all the variables which will be output for
%further analysis
output1=timestep*[tab1(1:length(tab2)) tab2(1:length(tab2)) gdinterval(:)
airinterval(:) totalperiod(:)];%converts indices to times
output2=timestep*pkforceindex(1:length(tab2));%converts index to time
output3=pkforcevalue(1:length(tab2));%peak force value
output5=1./(timestep*totalperiod(:));%bounce frequency

%[force start time|force end time|length of ground interval|length
%of air interval|total bounce period|peak force time|peak
%force|bounce frequency]
output=[output1 output2 output3 output5];

gdpercent=(output(:,3)./output(:,5))*100;%percent of interval on ground
airpercent=(output(:,4)./output(:,5))*100;%percent of interval in air
pkforcep=((output(:,6)-output(:,1))./output(:,5))*100;%percent of interval where
peak force occurs
output4=[gdpercent airpercent pkforcep];%[on ground|in air|when peak force
occurs]
output6=(1/100)*pkforcep.*output1(:,5);%[time into bounce cycle when peak force
occurs]

slope1=output3./output6;%slope touchdown to peakforce
timefpktoto=(output1(:,3)-output6);%time from peak force to toe off
slope2=output3./timefpktoto;%slope peak force to toe off
pkforcen1=(1/108)*output3;%peak force normalized to baby weight

if 1<=z<=2
    pkforcen2=(1/137.34)*output3;%peak force normalized to system weight
elseif z>=6
    pkforcen2=(1/127.53)*output3;%peak force normalized to system weight
else
    pkforcen2=(1/156.96)*output3;
end
output7=[slope1 slope2 timefpktoto pkforcen1 pkforcen2];

%fulloutput=[A force start time|B force end time|C length of ground interval|D
length
%of air interval|E total bounce period|F peak force time|G peak
%force|H bounce frequency|I percent on ground|J percent in air|K percent when peak
force
%occurs|L time into bounce cycle when peak force occurs|
%M slope touchdown to peakforce |N slope peak force to toe off |
%O time from peak force to toe off|P peak force normalized to baby weight|
%Q peak force normalized to system weight]
fulloutput=[0 0 0 0 0 0 0 0 0 0 0 0 0 0 0; output output4 output6 output7];

csvwrite(strcat('C',num2str(q),'T',num2str(r),num2str(s),'F.csv'),fulloutput)
end
clear all

end

```

Appendix C

- File *importfile4.m* imports filtered kinematic data and timing for appendix B and C
- File *processbouncekinematics.m* MATLAB script for the kinematic dependent variables.

```
function importfile4(fileToRead1)
%IMPORTFILE4(FILETOREAD1)
% Imports data from the specified file
% FILETOREAD1: file to read

% Auto-generated by MATLAB on 26-May-2010 23:42:59

DELIMITER = ',';
HEADERLINES = 1;

% Import the file
newData1 = importdata(fileToRead1, DELIMITER, HEADERLINES);

% Create new variables in the base workspace from those fields.
vars = fieldnames(newData1);
for i = 1:length(vars)
    assignin('base', vars{i}, newData1.(vars{i}));
end

-----

%Outputs the following variables to a file with a label corresponding to
%the source file, but with the suffix of kin1.csv and kin2.csv May 27,2010

close all
clear all

%w is the number of condition trial pairs (number of rows in the variables CT and trial)
for w=1:11

    %CT represents the condition (1 through 4), trial number, and number of
    %bouts per specified trial
    CT=[1 2 8;1 3 9;2 5 11;2 6 5;2 7 1;2 8 2;3 2 1;3 3 13;3 4 9;4 5 14;4 6 5];

    %the rows in the variable trial represent the bouts in the trial specified
    %by the condition and trial number in the same row of the matrix CT
    trial=[3 4 6 8 9 10 11 12 0 0 0 0 0 0;
          1 2 3 4 5 6 7 8 9 0 0 0 0 0;
          1 2 4 5 6 7 8 9 10 11 12 0 0 0;
          1 2 3 5 6 0 0 0 0 0 0 0 0 0;
          1 0 0 0 0 0 0 0 0 0 0 0 0 0;
          1 2 0 0 0 0 0 0 0 0 0 0 0 0;
          1 0 0 0 0 0 0 0 0 0 0 0 0 0;
          1 2 4 5 7 8 9 11 12 14 16 18 19 0;
          2 3 4 5 6 7 8 9 11 0 0 0 0 0;
          1 2 3 4 5 6 7 8 9 10 11 12 13 14;
          1 2 3 4 5 0 0 0 0 0 0 0 0 0];

    q=CT(w,1);%condition number
    r=CT(w,2);%trial number
    counter=0;

    for y=1:CT(w,3) %number of bouts
        %calls the appropriate file of raw data for the condition,trial and
        %bout specified
        s=trial(w,y);

        importfile2(strcat('IS05C',num2str(q),'T',num2str(r),num2str(s),'.txt'))
        datac=data;
        importfile4(strcat('C',num2str(q),'T',num2str(r),num2str(s),'F.csv'))
        data2=data;

        freq=120; %in Hz

        %force start and force end
```

```

forcestart=data2(:,1);%column containing the start time of each bounce cycle
forceend=data2(:,2);%column containing the end time of ground interval of the
cycle
pkforcetime=data2(:,6);%column containing the time at which the peak force occurs
in each cycle
forcestartindex=round(forcestart./(1/freq));%column containing the index number at
the start of a bounce cycle
forceendindex=round(forceend./(1/freq));%column containing the index number at the
end of each bounce cycle
pkforceindex=round(pkforcetime./(1/freq));%column containing the index number at
the peak force in each bounce cycle

%joint data from data set
ra=atac(:,9); %right ankle
la=atac(:,3); %left ankle
rk=atac(:,11); %right knee
lk=atac(:,5); %left knee
rh=atac(:,13); %right hip
lh=atac(:,7); %left hip

bounceheight=atac(:,22); %height of the t8 marker

joint=[ra(:) la(:) rk(:) lk(:) rh(:) lh(:)]';

%initializes matrices for collecting data in cycles
matrix=ones(500,7*length(forcestart));
matrixb=ones(500,7*length(forcestart));
matrixc=ones(500,7*length(forcestart));

for z=1:6 %once for each joint
    %to find angular position at specific force times
    for x=1:length(forcestart)-1 %once for each bounce cycle
        start=forcestartindex(x,1); %start of bounce cycle x
        nextstart=forcestartindex(x+1,1);%end of bounce
        cend=forceendindex(x,1); %end of ground interval in bounce cycle x
        dnm=nextstart-1-start; %range of bounce cycle x
        denom=dnm/freq;
        pk=pkforceindex(x,1); %peak force within bounce cycle x

        counter=counter+1;

        cycle=[joint(z,start:nextstart-1)]; %joint(determined by z) data
        for bounce cycle x [row vector]
            cyc=cycle'; %transforms to column vector
            bhcycle=bounceheight(start:nextstart-1,1); %height of the t8
            marker during cycle x

            for p=1:length(cyc)%once for each time point in cycle x
                c1(p,1)=100*(p/dnm); %converts to percentage of bounce cycle x
                c2(p,1)=cyc(p,:); %value of joint angle at that
                time/percentage point
                c3(p,1)=bhcycle(p,:); %height of T8 at that time/percentage
                point
            end

            %interval of the bounce cycle where x indicates which bounce
            %cycle
            footcontact(x,z)=joint(z,start); %angle at the time of foot
            contacting the ground, i.e., start of cycle
            toeff(x,z)=joint(z,cend); %angle at time when foot leaves the
            ground

            pkforcecy(x,z)=joint(z,pk);%peak force within bounce cycle

            %to find intervals of increase or decrease of
            %bounceheight
            bicounter=0;
            bdcouter=0;

        [bouncemax(x,1),bouncemaxindex(x,1)]=max(bounceheight(start:nextstart-1,1));
        [bouncemin(x,1),bounceminindex(x,1)]=min(bounceheight(start:nextstart-1,1));
        for u=1:length(bhcycle)-1

```

```

        if bhcycle(u+1,1)-bhcycle(u,1)>=0
            %increasing
            bicounter=bicounter+1;
        else
            %decreasing
            bdcounter=bdcounter+1;
        end
    end
    biinterval(x,1)=bicounter/freq; %interval of increasing T8 height
    biintervalp(x,1)=biinterval(x,1)/denom;%percent of cycle interval
of increasing T8 height
    bdinterval(x,1)=bdcounter/freq; %interval of decreasing T8 height
    bdintervalp(x,1)=bdinterval(x,1)/denom;%percent of cycle interval
of decreasing T8 height
    [maxjointangle(x,z),maxjointindex(x,z)]=max(cycle);%max joint
angle within bounce cycle, index of maximum joint angle
    [minjointangle(x,z),minjointindex(x,z)]=min(cycle);%min joint
angle within bounce cycle, index of minimum joint angle
    maxjapercent(x,z)=maxjointindex(x,z)/(nextstart-start);%percent
of cycle x at which the maximum joint angle occurs
    minjapercent(x,z)=minjointindex(x,z)/(nextstart-start);%percent
of cycle x at which the minimum joint angle occurs
    AP1(z,x)=joint(z,bouncemaxindex(x,1));%angles of joints at max T8
height
    AP2(z,x)=joint(z,bounceminindex(x,1));%angles of joints at min T8
height

    maxjointindexg(x,z)=maxjointindex(x,z)+start;
    minjointindexg(x,z)=minjointindex(x,z)+start;
    bounceminindexg(x,1)=bounceminindex(x,1)+start;

    %to find intervals of increase or decrease in joint
    %angles
    icounter=0;
    dcounter=0;
    for v=1:length(cyc)-1
        if cyc(v+1,1)-cyc(v,1)>=0
            %increasing
            icounter=icounter+1;
        else
            %decreasing
            dcounter=dcounter+1;
        end
    end
    iinterval(x,z)=icounter/freq;%interval of joint extension
    iintervalp(x,z)=iinterval(x,z)/denom;%percent of cycle interval of
extension
    dinterval(x,z)=dcounter/freq;%interval of joint flexion
    dintervalp(x,z)=dinterval(x,z)/denom;%percent of cycle interval of
flexion

    matrix(1:length(c1),7*x-6)=c1;
    matrix(1:length(c2),7*x-(6-z))=c2;
    matrixb(1:length(c3),x)=c3;
    matrixc(1:length(c1),x)=c1;

    clear cycle c1 c2 c3
end
end

    output=[footcontact tocoeff pkforcecyc maxjointangle maxjapercent minjointangle
minjapercent bouncemax bouncemin];
    output2=[bdinterval bdintervalp biinterval biintervalp iinterval iintervalp
dinterval dintervalp AP1' AP2'];
    output3=[0 0 0 0 0 0 0 0 0 0 0 0; maxjointindexg minjointindexg
bounceminindexg];

    clear footcontact tocoeff pkforcecyc maxjointangle maxjapercent minjointangle
minjapercent bouncemax bouncemin
    clear bdinterval bdintervalp biinterval biintervalp iinterval iintervalp dinterval
dintervalp AP1 AP2
    clear maxjointindexg minjointindexg bounceminindexg

```

```

csvwrite(strcat('C',num2str(q),'T',num2str(r),num2str(s),'kin1.csv'),output)
%A-F angles of the joints at the time of foot contact
%G-L angles of the joints at the time of toe off
%M-R angles of the joints at the time of peak force during the cycle
%S-X Maximum joint angles during the cycle
%Y-AD Percent at which maximum joint angles occur
%AE-AJ Minimum joint angles during the cycle
%AK-AP Percent at which minimum joint angles occur
%AQ Maximum T8 height
%AR Minimum T8 height

csvwrite(strcat('C',num2str(q),'T',num2str(r),num2str(s),'kin2.csv'),output2)
%A Time T8 height is decreasing
%B Percent of cycle where T8 height is decreasing
%C Time T8 height is decreasing
%D Percent of cycle where T8 height is decreasing
%E-J Time in extension
%K-P Percent of cycle in extension
%Q-V Time in flexion
%W-AB Percent of cycle in flexion
%AC-AH angles of joints at maximum T8 height
%AI-AN angles of joints at minimum T8 height

csvwrite(strcat('C',num2str(q),'T',num2str(r),num2str(s),'mjmj.csv'),output3)
%A-F index (kin) when maximum joint angles occur
%G-L index (kin) when minimum joint angles occur
%M index (kin) when minimum T8 height occurs

co=(counter/6);

for k=7:7:7*x
    mja1(:,2*(k/7+co)-1)=matrix(:,k-6);
    mja1(:,2*(k/7+co))=matrix(:,k-5);
    mja2(:,2*(k/7+co)-1)=matrix(:,k-6);
    mja2(:,2*(k/7+co))=matrix(:,k-4);
    mja3(:,2*(k/7+co)-1)=matrix(:,k-6);
    mja3(:,2*(k/7+co))=matrix(:,k-3);
    mja4(:,2*(k/7+co)-1)=matrix(:,k-6);
    mja4(:,2*(k/7+co))=matrix(:,k-2);
    mja5(:,2*(k/7+co)-1)=matrix(:,k-6);
    mja5(:,2*(k/7+co))=matrix(:,k-1);
    mja6(:,2*(k/7+co)-1)=matrix(:,k-6);
    mja6(:,2*(k/7+co))=matrix(:,k);
    mjab(:,k/7+co)=matrixc(:,k/7);
    mjab(:,k/7+1+co)=matrixb(:,k/7);
end

clear matrix matrixb matrixc
end

if w==2
    xlswrite('condition1.xlsx',mja1,'rightankle')
    xlswrite('condition1.xlsx',mja2,'leftankle')
    xlswrite('condition1.xlsx',mja3,'rightknee')
    xlswrite('condition1.xlsx',mja4,'leftknee')
    xlswrite('condition1.xlsx',mja5,'righthip')
    xlswrite('condition1.xlsx',mja6,'lefthip')
    xlswrite('condition1.xlsx',mjab,'bounceheight')
    clear all
elseif w==6
    xlswrite('condition2.xlsx',mja1,'rightankle')
    xlswrite('condition2.xlsx',mja2,'leftankle')
    xlswrite('condition2.xlsx',mja3,'rightknee')
    xlswrite('condition2.xlsx',mja4,'leftknee')
    xlswrite('condition2.xlsx',mja5,'righthip')
    xlswrite('condition2.xlsx',mja6,'lefthip')
    xlswrite('condition2.xlsx',mjab,'bounceheight')
    clear all
elseif w==9
    xlswrite('condition3.xlsx',mja1,'rightankle')
    xlswrite('condition3.xlsx',mja2,'leftankle')

```

```

        xlswrite('condition3.xlsx',mja3,'rightknee')
        xlswrite('condition3.xlsx',mja4,'leftknee')
        xlswrite('condition3.xlsx',mja5,'righthip')
        xlswrite('condition3.xlsx',mja6,'lefthip')
        xlswrite('condition3.xlsx',mjab,'bounceheight')
        clear all
elseif w==11
    xlswrite('condition4.xlsx',mja1,'rightankle')
    xlswrite('condition4.xlsx',mja2,'leftankle')
    xlswrite('condition4.xlsx',mja3,'rightknee')
    xlswrite('condition4.xlsx',mja4,'leftknee')
    xlswrite('condition4.xlsx',mja5,'righthip')
    xlswrite('condition4.xlsx',mja6,'lefthip')
    xlswrite('condition4.xlsx',mjab,'bounceheight')
    clear all
else
    h=1;
end
clearvars -except mja1 mja2 mja3 mja4 mja5 mja6 mjab counter
end
end

```

Appendix D

- File *processjointextrema.m* MATLAB script for the kinematic data at specific bounce cycle events.

```

%Outputs the following variables to a file with a label corresponding to
%the source file, but with the suffix of M.csv

%Variables:
    %[A-F vertical ground reaction force at maximum joint angles|G-L vertical ground
reaction force at minimum joint angles|M vertical ground reaction force at the minimum T8
height]

close all
clear all

%z is the number of condition trial pairs (number of rows in the variables CT and trial)
for z=1:11

    %CT represents the condition (1 through 4), trial number, and number of
    %bouts per specified trial
    CT=[1 2 8;1 3 9;2 5 11;2 6 5;2 7 1;2 8 2;3 2 1;3 3 13;3 4 9;4 5 14;4 6 5];

    %the rows in the variable trial represent the bouts in the trial specified
    %by the condition and trial number in the same row of the matrix CT
    trial=[3 4 6 8 9 10 11 12 0 0 0 0 0 0;
        1 2 3 4 5 6 7 8 9 0 0 0 0 0;
        1 2 4 5 6 7 8 9 10 11 12 0 0 0;
        1 2 3 5 6 0 0 0 0 0 0 0 0 0;
        1 0 0 0 0 0 0 0 0 0 0 0 0 0;
        1 2 0 0 0 0 0 0 0 0 0 0 0 0;
        1 0 0 0 0 0 0 0 0 0 0 0 0 0;
        1 2 4 5 7 8 9 11 12 14 16 18 19 0;
        2 3 4 5 6 7 8 9 11 0 0 0 0 0;
        1 2 3 4 5 6 7 8 9 10 11 12 13 14;
        1 2 3 4 5 0 0 0 0 0 0 0 0 0];

    q=CT(z,1);%condition number
    r=CT(z,2);%trial number

    for y=1:CT(z,3) %number of bouts
        %calls the appropriate file of raw data for the condition,trial, and
        %bout specified
        s=trial(z,y);
        importfile2(strcat('IS05C',num2str(q),'T',num2str(r),num2str(s),'F.txt'))
        forcedata=[data(:,1) data(:,4)];%index number and total force
        importfile4(strcat('C',num2str(q),'T',num2str(r),num2str(s),'mjmj.csv'))
        maxind=5*data(:,1:6);
        minind=5*data(:,7:12);
        bouncemin=data(:,13);

        for u=1:length(bouncemin)
            fatbouncemin(u,1)=forcedata(bouncemin(u,1),2);
            for v=1:6
                fatmaxflex(u,v)=forcedata(maxind(u,v),2);
                fatminext(u,v)=forcedata(minind(u,v),2);
            end
        end

        output=[fatmaxflex fatminext fatbouncemin];
        %[A-F vertical ground reaction force at maximum joint angles|G-L
        %vertical ground reaction force at minimum joint angles|M vertical ground reaction
force at the minimum T8 height]
        csvwrite(strcat('C',num2str(q),'T',num2str(r),num2str(s),'M.csv'),output)
        clear maxind minind bouncemin fatbouncemin fatminext fatmaxflex
    end
clear all
end

```

Appendix E

- File *importfile.m* imports filtered kinematic data for appendix D, E and F
- File *bouncecycles#* denotes a MATLAB script to extract normalized kinematic data for the left and right ankle, knee and hip, and the T8 position.

```
function importbfile(fileToRead1)
%IMPORTBFILE(FILETOREAD1)
% Imports data from the specified file
% FILETOREAD1: file to read

% Auto-generated by MATLAB on 17-Jul-2010 15:05:39

% Import the file
rawData1 = importdata(fileToRead1);

% For some simple files (such as a CSV or JPEG files), IMPORTDATA might
% return a simple array. If so, generate a structure so that the output
% matches that from the Import Wizard.
[unused,name] = fileparts(fileToRead1);
newData1.(genvarname(name)) = rawData1;

% Create new variables in the base workspace from those fields.
vars = fieldnames(newData1);
for i = 1:length(vars)
    assignin('base', vars{i}, newData1.(vars{i}));
end

-----

clear all
close all

b=1;
importbfile(strcat('c',num2str(b),'bounces.csv'))
bdata=c\bounces;

c=2;
for a=1:length(bdata)-1
    if bdata(a+1,1)<=bdata(a,1)
        newbout(1,1)=0;
        newbout(c,1)=a;
        c=c+1;
    else
        end
end

freq=120; %in Hz
counter=2;

for d=1:length(newbout)-1
    if (0<d && d<=8)
        CT=[1 2 8];
        trial=[3 4 6 8 9 10 11 12 0 0 0 0 0 0];
        r=CT(2);%trial number
        s=trial(d);
        importfile2(strcat('IS05C',num2str(b),'T',num2str(r),num2str(s),'.txt'))
        datac=data;
    else
        CT=[1 3 9];
        trial=[1 2 3 4 5 6 7 8 9 0 0 0 0 0];
        r=CT(2);%trial number
        s=trial(d-8);
        importfile2(strcat('IS05C',num2str(b),'T',num2str(r),num2str(s),'.txt'))
        datac=data;
    end

    clear bd
    bd=bdata((newbout(d)+1):newbout(d+1),:);
```

```

%joint data from data set
ra=datac(:,9); %right ankle
la=datac(:,3); %left ankle
rk=datac(:,11); %right knee
lk=datac(:,5); %left knee
rh=datac(:,13); %right hip
lh=datac(:,7); %left hip

bounceheight=datac(:,22); %height of the t8 marker

%ranges
for f=1:length(bd)
    cyclestartindex=round(freq*bd(f,1));
    cyclelength=round(freq*bd(f,2));
    cycleendindex=cyclestartindex+cyclelength-1;

    clear rab lab rkb lkb rhb lhb bhb c1 tsrab

    rab=ra(cyclestartindex:cycleendindex,1);
    lab=la(cyclestartindex:cycleendindex,1);
    rkb=rk(cyclestartindex:cycleendindex,1);
    lkb=lk(cyclestartindex:cycleendindex,1);
    rhb=rh(cyclestartindex:cycleendindex,1);
    lhb=lh(cyclestartindex:cycleendindex,1);
    bhb=bounceheight(cyclestartindex:cycleendindex,1);

    for p=1:cyclelength %once for each time point in cycle x
        c1(p,1)=100*(p/cyclelength); %converts to percentage of bounce cycle x
    end

    p1=[0:4:100]';

    %resamples to the same length, 0 to 100 increasing by 4
    tsrab=timeseries(rab,c1);
    rabp=resample(tsrab,p1);

    tslab=timeseries(lab,c1);
    labp=resample(tslab,p1);

    tsrkb=timeseries(rkb,c1);
    rkbp=resample(tsrkb,p1);

    tslkb=timeseries(lkb,c1);
    lkbp=resample(tslkb,p1);

    tsrhb=timeseries(rhb,c1);
    rhbp=resample(tsrhb,p1);

    tslhb=timeseries(lhb,c1);
    lhbp=resample(tslhb,p1);

    tsbhb=timeseries(bhb,c1);
    bhbp=resample(tsbhb,p1);

    %bounce cycles to be output
    rightankle(:,counter)=rabp.Data;
    leftankle(:,counter)=labp.Data;
    rightknee(:,counter)=rkbp.Data;
    leftknee(:,counter)=lkbp.Data;
    righthip(:,counter)=rhbp.Data;
    lefthip(:,counter)=lhbp.Data;
    t8height(:,counter)=bhbp.Data;

    rightankle(1,counter)=ra(cyclestartindex,1);
    leftankle(1,counter)=la(cyclestartindex,1);
    rightknee(1,counter)=rk(cyclestartindex,1);
    leftknee(1,counter)=lk(cyclestartindex,1);
    righthip(1,counter)=rh(cyclestartindex,1);
    lefthip(1,counter)=lh(cyclestartindex,1);
    t8height(1,counter)=bounceheight(cyclestartindex,1);

    counter=counter+1;
end

```

end

```
rightankle(:,1)=p1;
leftankle(:,1)=p1;
rightknee(:,1)=p1;
leftknee(:,1)=p1;
righthip(:,1)=p1;
lefthip(:,1)=p1;
t8height(:,1)=p1;

csvwrite('conditionlrabounces.csv',rightankle)
csvwrite('conditionllabounces.csv',leftankle)
csvwrite('conditionlrkbounces.csv',rightknee)
csvwrite('conditionllkbounces.csv',leftknee)
csvwrite('conditionlrbounces.csv',righthip)
csvwrite('conditionllhbounces.csv',lefthip)
csvwrite('conditionlt8bounces.csv',t8height)
```

Appendix F

- File *forcecycles#* denotes a MATLAB script to extract normalized vertical ground reaction forces data.

```
clear all
close all

b=1;
importbfile(strcat('c',num2str(b),'bounces.csv'))
bdata=c1bounces;

c=2;
for a=1:length(bdata)-1
    if bdata(a+1,1)<=bdata(a,1)
        newbout(1,1)=0;
        newbout(c,1)=a;
        c=c+1;
    else
        end
end

freq=600; %in Hz
counter=2;

for d=1:length(newbout)-1
    if (0<d && d<=8)
        CT=[1 2 8];
        trial=[3 4 6 8 9 10 11 12 0 0 0 0 0 0];
        r=CT(2);%trial number
        s=trial(d);
        importfile2(strcat('IS05C',num2str(b),'T',num2str(r),num2str(s),'F.txt'))
        datac=data;
    else
        CT=[1 3 9];
        trial=[1 2 3 4 5 6 7 8 9 0 0 0 0 0];
        r=CT(2);%trial number
        s=trial(d-8);
        importfile2(strcat('IS05C',num2str(b),'T',num2str(r),num2str(s),'F.txt'))
        datac=data;
    end

    clear bd
    bd=bdata((newbout(d)+1):newbout(d+1),:);

    %joint data from data set
    force=datac(:,4); %combined force

    %ranges
    for f=1:length(bd)
        cyclestartindex=round(freq*bd(f,1));
        cyclelength=round(freq*bd(f,2));
        cycleendindex=cyclestartindex+cyclelength-1;

        clear forceb c1 tsforce

        forceb=force(cyclestartindex:cycleendindex,1);

        for p=1:cyclelength %once for each time point in cycle x
            c1(p,1)=100*(p/cyclelength); %converts to percentage of bounce cycle x
        end

        p1=[0:4:100]';

        %resamples to the same length, 0 to 100 increasing by 4
        tsforce=timeseries(forceb,c1);
        forcebp=resample(tsforce,p1);
    end
end
```

```
    %bounce cycles to be output
    forcepeak(:,counter)=forcebp.Data;

    forcepeak(1,counter)=force(cyclestartindex,1);
    counter=counter+1;
end

end

forcepeak(:,1)=p1;

csvwrite('condition1forcepeak.csv',forcepeak)
```

Appendix G

– File *c#bcnonnormalized.m* denotes a MATLAB script that exports non-normalized kinematic data for the left and right ankle, knee and hip.

```
clear all
close all

b=1;
importbfile(strcat('c',num2str(b),'bounces.csv'))
bdata=c1bounces;

c=2;
for a=1:length(bdata)-1
    if bdata(a+1,1)<=bdata(a,1)
        newbout(1,1)=0;
        newbout(c,1)=a;
        c=c+1;
    else
    end
end

freq=120; %in Hz
counter=2;
time=[0:1/120:149/120];

for d=1:length(newbout)-1
    if (0<d && d<=8)
        CT=[1 2 8];
        trial=[3 4 6 8 9 10 11 12 0 0 0 0 0];
        r=CT(2);%trial number
        s=trial(d);
        importfile2(strcat('IS05C',num2str(b),'T',num2str(r),num2str(s),'.txt'))
        datac=data;
    else
        CT=[1 3 9];
        trial=[1 2 3 4 5 6 7 8 9 0 0 0 0];
        r=CT(2);%trial number
        s=trial(d-8);
        importfile2(strcat('IS05C',num2str(b),'T',num2str(r),num2str(s),'.txt'))
        datac=data;
    end

    clear bd
    bd=bdata((newbout(d)+1):newbout(d+1),:);

    %joint data from data set
    ra=datac(:,9); %right ankle angular position
    la=datac(:,3); %left ankle angular position
    rk=datac(:,11); %right knee angular position
    lk=datac(:,5); %left knee angular position
    rh=datac(:,13); %right hip angular position
    lh=datac(:,7); %left hip angular position

    %ranges
    for f=1:length(bd)
        cyclestartindex=round(freq*bd(f,1));
        cyclelength=round(freq*bd(f,2));
        cycleendindex=cyclestartindex+cyclelength-1;

        clear rab lab rkb lkb rhb lhb

        rab=ra(cyclestartindex:cycleendindex,1);
        lab=la(cyclestartindex:cycleendindex,1);
        rkb=rk(cyclestartindex:cycleendindex,1);
        lkb=lk(cyclestartindex:cycleendindex,1);
        rhb=rh(cyclestartindex:cycleendindex,1);
        lhb=lh(cyclestartindex:cycleendindex,1);
```

```

for h=1:150
    if h<=length(rab)
        apra(h,counter)=rab(h,1);
        apla(h,counter)=lab(h,1);
        aprk(h,counter)=rkb(h,1);
        aplk(h,counter)=lkb(h,1);
        aprh(h,counter)=rhb(h,1);
        aplh(h,counter)=lhb(h,1);
    else
        apra(h,counter)=nan(1,1);
        apla(h,counter)=nan(1,1);
        aprk(h,counter)=nan(1,1);
        aplk(h,counter)=nan(1,1);
        aprh(h,counter)=nan(1,1);
        aplh(h,counter)=nan(1,1);
    end
end
    counter=counter+1;
end

end

apra(:,1)=time';
apla(:,1)=time';
aprk(:,1)=time';
aplk(:,1)=time';
aprh(:,1)=time';
aplh(:,1)=time';

csvwrite('clrightanklenn.csv', apra)
csvwrite('clleftanklenn.csv', apla)
csvwrite('clrightkneenn.csv', aprk)
csvwrite('clleftkneenn.csv', aplk)
csvwrite('clrighthipnn.csv', aprh)
csvwrite('cllefthipnn.csv', aplh)

```

Appendix H

Letter of Information and Consents

INFORMED CONSENT TO PARTICIPATE IN A STUDY ENTITLED: DEVELOPMENT OF INFANT BOUNCING BEHAVIOUR

I, _____, have been invited to allow the participation of my infant _____ in the research study entitled **DEVELOPMENT OF INFANT BOUNCING BEHAVIOUR**.

Researchers: Heidi Sveistrup, Gordon Robertson

OBJECTIVE OF THE RESEARCH STUDY

The primary objective is to characterize body movement in infants bouncing in a harness mounted on a spring system. We are interested in determining whether infants use different patterns as they bounce. We are also interested in determining if infants will adapt to their external environments and change their bouncing patterns. Consequently, we will understand the steps or pattern of human development when learning a new task and how infants adapt when you change the environment.

PARTICIPATION IN THE STUDY

My participation will require that my infant attend two two-hour sessions at the University of Ottawa.

At the start of the session, I will complete a brief questionnaire about my infant's health. This will take approximately five minutes. My infant will then be undressed (to his/her diaper and shorts) and small reflective balls will be attached to his/her skin with hypoallergenic tape. The balls will be placed on the side of the feet, ankles, knees, hips, waist and shoulders as well as on the chest, and back of my infant. My infant will also wear a headband that will have additional balls attached to it. My infant will then be placed in a harness attached to a spring system and encouraged to bounce. Cameras will be used to record the position of the reflective balls that are placed on my child. The cameras used do not record images of my infant. The ball positions will be used to reconstruct my infant's behaviour. My child's bouncing will be recorded for up to 30 minutes. In addition to marker recording, the forces that my child produces while jumping will be recorded using a weight scale mounted into the floor. During the testing period, rest periods will be scheduled so that my child will not become fatigued. If I feel my child needs additional rest, I can simply indicate this to the experimenter who will provide additional rest breaks.

Following the testing session, the experimenter will observe my infant and record his/her behavior for approximately 20 minutes. During this time, my infant will be videotaped for subsequent analysis. Finally, my child's weight, height, limb and joint measurements will be taken. The session will take no longer than two hours.

In order to participate in this study, I understand that my child must not have had any recent or prior medical care beyond regular pediatric visits.

RISKS

There are minimal risks to my infant associated with my participation in this study. There is a risk that my child may become fatigued during test or training sessions. Rest periods have been scheduled into the sessions to reduce fatigue and I can indicate if additional rest is needed.

BENEFITS OF PARTICIPATING IN THIS STUDY

There are no direct benefits to me or to my infant for participating in this study. The data from this study will help increase our understanding of how infants control and use their legs to generate very complex behaviours. The data may also provide us with a normative dataset that we can use to better understand movement control in infants who are unable to bounce due to disability.

COMPENSATION

There is no compensation for participating in this study. However, costs for travel to and from the Research Centre for test sessions will be reimbursed by a maximum amount of \$20.00.

CONFIDENTIALITY AND ANONYMITY

Information from the testing and training sessions will be held in strictest confidence and will not be used except for research purposes. It is understood that my infant's file and any data from my infant will be coded in a way that will ensure anonymity. I understand that any published results will be presented with complete anonymity. I understand that any personal data (e.g., name, address, telephone number) will be kept in a separate file folder, accessible only by the two researchers and a research assistant responsible for the project. All data will be kept for five years after publication, in a locked filing cabinet in an office at the University of Ottawa and will be destroyed at the end of this period.

QUESTIONS CONCERNING THIS STUDY

I understand that I may ask all my questions regarding this study and they will be answered. These questions may be addressed to Dr. Heidi Sveistrup or Dr. Gordon Robertson, project investigators.

If I have any questions with respect to the ethical conduct of this study, I may contact the Protocol Officer for Ethics in Research, University of Ottawa, Tabaret Hall.

VOLUNTARY PARTICIPATION

I understand that participation is voluntary, and that I may withdraw my infant at any time for any reason, without prejudice. In the event of withdrawal, all of the information gathered will be destroyed and will not be used.

CONSENT

I declare that I understand this project, the nature and the degree of my infant’s participation and the possible disadvantages and risks of the project as listed in this consent form. I have had the opportunity to ask all my questions concerning the different aspects of the study and have received responses to my satisfaction.

I _____ the undersigned, voluntarily accept to have my infant participate in this study. I can withdraw my infant from the study at any time without any prejudice. I certify that I have had adequate time to make my decision.

There are two signed copies of this consent form, one of which is for me to keep.

Name of parent

Signature of parent

Date

Name of investigator

Signature of investigator

Date

University of Ottawa, Faculty of Health Sciences

Subject ID: _____

DEVELOPMENT OF INFANT BOUNCING BEHAVIOUR

USE OF VIDEO FOR DATA ANALYSIS PURPOSES

Name of researchers: Habib Perez, Olinda, MSc Candidate
Dr. Heidi Sveistrup
Dr. D.G.E. Robertson

Institution, Faculty, Department: University of Ottawa, Faculty of Health Sciences,
School of Human Kinetics and Rehabilitation Sciences

E-mail address:

Videos will only be viewed and used by the research team listed above and will be kept in our records for a maximum duration of five years.

Please initial one of the items below:

I **agree** to have my infant videotaped during the Alberta Infant Motor Scale (AIMS) for subsequent analysis. _____ (initials)

I **do not agree** to have my infant videotaped during the Alberta Infant Motor Scale (AIMS) for subsequent analysis. _____ (initials)

If I have agreed that the videos be used for subsequent analysis purposes and I later decide to withdraw such consent, all I need to do is contact the researcher or supervisors at the numbers mentioned above, and the videos will no longer be used.

Parent/Guardian's signature: _____

Date: _____

INFANT DATA SHEET

TESTING DATE (DD/MM/YY):	
SUBJECT NUMBER/ID:	SEX: M F
DATE OF BIRTH (DD/MM/YY):	If not to term, No. of months to term:
AGE AT ASSEMENT:	
JJ AT HOME? Y / N	No. of HOURS IN JJ AT HOME/week:
Has the baby seen a doctor for anything else than regular check-ups? Y / N	
AIMS Score:	PERCENTILE:

BODY PARAMETERS

GENERAL

Height (cm)	Weight (kg/lbs)
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SEGMENT LENGHTS (cm)

(Note: all segment lengths will be measured between the corresponding markers for that segment)

THIGH (From greater trochanter to lateral epicondyle of the knee and corresponding axis of rotation)

LEFT:	RIGHT:
-------	--------

SHANK (From lateral epicondyle of the knee and the corresponding axis of rotation to the lateral malleolus)

LEFT:	RIGHT:
-------	--------

FOOT (From the lateral malleolus to ball of foot)

LEFT:	RIGHT:
-------	--------

SEGMENT GIRTHS (cm)

THIGH (Circumference at thickest point of the thigh segment)

SHANK (Circumference at thickest point of the shank segment.)

LEFT:	RIGHT:	LEFT:	RIGHT:
-------	--------	-------	--------

FEET (Width of the foot at its widest part/ ball of foot)

LEFT:	RIGHT:
-------	--------

JOINT WIDTHS (cm)

KNEE (Horizontal width from Medial Menisci to Lateral Menisci)

ANKLE (Horizontal width from Medial Malleolus to Lateral Malleolus)

LEFT:	RIGHT:	LEFT:	RIGHT:
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Appendix I

Ethics Approval Notice

File Number: H04-09-07

Date (mm/dd/yyyy): 06/24/2009



Université d'Ottawa **University of Ottawa**
Service de subventions de recherche et déontologie Research Grants and Ethics Services

Ethics Approval Notice

Health Sciences and Science REB

Principal Investigator / Supervisor / Co-investigator(s) / Student(s)

<u>First Name</u>	<u>Last Name</u>	<u>Affiliation</u>	<u>Role</u>
Heidi	Sveistrup	Health Sciences / Occupational Therapy	Principal Investigator
D. Gordon E.	Robertson	Health Sciences / Human Kinetics	Co-investigator

File Number: H04-09-07

Type of Project: Professor

Title: Development of Infant Bouncing Behaviour

Approval Date (mm/dd/yyyy)	Expiry Date (mm/dd/yyyy)	Approval Type
06/24/2009	06/23/2010	Ia

(Ia: Approval, Ib: Approval for initial stage only)

Special Conditions / Comments:

N/A

1

550, rue Cumberland 550 Cumberland Street
Ottawa (Ontario) K1N 6N5 Canada Ottawa, Ontario K1N 6N5 Canada
(613) 562-5841 • Téléc./Fax (613) 562-5338
<http://www.rges.uottawa.ca> <http://www.ssr.uottawa.ca>



Université d'Ottawa **University of Ottawa**
Service de subventions de recherche et déontologie Research Grants and Ethics Services

This is to confirm that the University of Ottawa Research Ethics Board identified above, which operates in accordance with the Tri-Council Policy Statement and other applicable laws and regulations in Ontario, has examined and approved the application for ethical approval for the above named research project as of the Ethics Approval Date indicated for the period above and subject to the conditions listed the section above entitled "Special Conditions / Comments".

During the course of the study the protocol may not be modified without prior written approval from the REB except when necessary to remove subjects from immediate endangerment or when the modification(s) pertain to only administrative or logistical components of the study (e.g. change of telephone number). Investigators must also promptly alert the REB of any changes which increase the risk to participant(s), any changes which considerably affect the conduct of the project, all unanticipated and harmful events that occur, and new information that may negatively affect the conduct of the project and safety of the participant(s). Modifications to the project, information/consent documentation, and/or recruitment documentation, should be submitted to this office for approval using the "Modification to research project" form available at:
http://www.rges.uottawa.ca/ethics/application_dwn.asp

Please submit an annual status report to the Protocol Officer 4 weeks before the above-referenced expiry date to either close the file or request a renewal of ethics approval. This document can be found at:
http://www.rges.uottawa.ca/ethics/application_dwn.asp

If you have any questions, please do not hesitate to contact the Ethics Office at extension or by e-mail at:

Germain Zongo
Assistant Director, Ethics (Interim)
For Dr. Daniel Lagarec, Chair of the Health Sciences and Sciences REB