Postural Control on Ice Hockey Skates

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Abstract

Stability on ice and while wearing skates is a more challenging task than compared to bare foot and on ground; however, no studies exist to quantify how postural responses may specifically differ between these two conditions. Thus the purpose of this study was to compare the postural control responses due to a surface perturbation between bare foot (BF) and skate (SK) wearing conditions as well as between elite and recreational level hockey players. Eleven elite and 13 recreational level hockey players completed 18 crouched standing ("hockey ready posture") trials for both BF and SK conditions. Static (30s) and dynamic perturbation tests (10 cm, 40cm/s, horizontal translations along each of the medio-lateral, and anterior-posterior axes) were conducted. A motion capture system recorded body kinematics and centre of mass (CoM), while two force plates measured the center of pressure (CoP). Overall, remarkably similar postural responses were observed between conditions. Interaction effects were noted; however they were small and generally less than 10 mm. Similarly, body kinematics yielded no statistical differences across conditions, with the exception of the knee response during the Anterior perturbation; where in it was determined that as a subjects' initial crouch position increased (i.e. greater than 40° hip flexion), their response

reversed from knee flexion to knee extension. In summary, the adoption of the crouched and wide foot stance is believed to have mitigated the external perturbations effect on instability. These findings have practical coaching implications. Further research is warranted to address, for example, upper body perturbations.

Abrégé

La stabilité sur patins à la glace est généralement considérée plus difficile comparée à pieds nus sur terre ferme. Aucune étude jusqu'à présent ne quantifie la stabilité posturale due à des perturbations sous ces conditions. L'objectif de cette étude était de comparer la stabilité posturale due à une perturbation entre les conditions pieds nus (BF) et patin (SK), et entre joueurs de hockey élites et récréatifs. Onze joueurs élites et 13 joueurs récréatifs ont complété 18 essais en position accroupie sous les deux conditions, pour des perturbations statiques (30s) et dynamiques (10cm, 40cm/s, translation horizontale, axe médial-latéral et antérieur-postérieur). La cinématique du corps et le centre de masse (CoM) ont été mesurés avec un système d'analyse du mouvement,et le centre de pression (CoP) a été mesuré avec 2 plateformes de force. Des résultats semblables ont été observés sous les deux conditions. Des interactions ont été trouvées, mais les différences étaient petites (généralement moins que 10 mm). Aucune différence statistique entre conditions n'a été observée pour la cinématique du corps, avec l'exception de l'effet dû à la perturbation antérieure sur le genou : lorsque la position du sujet était plus accroupie, il y avait une réaction d'extension plutôt que de flexion (pour une position plus étroite). En résumé, la position accroupie avec pieds écartés semble avoir atténué les effets de perturbations externes sur l'instabilité. Ces résultats ont des implications

pratiques pour l’entraînement. Des études futures sont nécessaires pour adresser, par exemple, les perturbations au haut du corps.

Chapter 1: Introduction and Review of Literature

1.01 Introduction to Postural Control

Skating is the most fundamental skill in ice hockey at any level of play. Skating technique has been studied in the past typically with the focus being placed on the kinematics or kinetics of the lower limbs during the forward stride. (Chang, Turcotte, Lefebvre, Montgomery, & Pearsall, 2002; De Koning, De Groot, & Ingen Schenau, 1991; Lafontaine, 2007; Upjohn, Turcotte, Pearsall, & Loh, 2008). However, little attention has been given to one of the most essential components of skating; balance and postural control.

In humans, balance refers to the complex dynamics of body posture used to prevent falling and is directly related to postural control (Winter, 1995a). In order to control one's posture and remain balanced in any activity, internal forces must be generated to negate any destabilizing forces, such as gravity, external perturbations, or limb movements (Horak & Macpherson, 2010). This force generation to regain balance is typically investigated through the mechanics of body movement, the position of the center of mass (CoM), and the location of the center of pressure (CoP). The CoM represents a single point on the human body equivalent to the total body mass, and is the weighted average of the mass of each body segment (Winter, 1995a). During quiet stance, the CoM is located roughly at the first sacral vertebra (Oatis, 2004) . The CoP is the location of the vertical ground reaction force, and represents a weighted average of all the pressures between the plantar side of the foot and the contact surface (Winter, 1995a). Although the CoP does not always coincide with the CoM, the acceleration of the CoM is continuously regulated by adjustments of the CoP (Winter, Prince, Frank, Powell, & Zabjek, 1996).

The human body is inherently an unstable structure, as two-thirds of one's body mass is located two-thirds above the ground. Consequently the body is often modelled as an inverted pendulum consisting of a narrow base of support (BoS) and larger upper body, pivoting about the ankles, the hips, or both (Winter, 1995b). In quiet upright stance, the CoM is located within one's BoS, which is the area defined by the body's feet and the support surface (Horak & Macpherson, 2010). The gravitational force acting on the CoM may generate torques resulting in angular accelerations of segments of the body. This force is continuously counter balanced by opposing ground reaction forces (GRF), through the CoP, creating counter torques. The slight time delays between these torques and counter torques result in body sway or oscillations about the ankle joint (Winter, 1995a). This ankle strategy during quiet upright stance represents postural control at the most basic level, and the challenge is minimal. However, if the BoS starts to decrease due to a narrower stance width, remaining balanced becomes much more difficult as the COM begins to approach the edge of one's

BoS (Henry, Fung, & Horak, 2001). The difficulty to remain balanced is further complicated if an external perturbation occurs or if the CoM moves outside of the BoS. This scenario may result in a fall unless more substantial postural control is executed through coordinated muscular contractions of the ankles and/or the hips.

Based on this model, in simple barefoot quiet stance the human body is mechanically unstable and requires constant control to remain balanced. Theoretically, if hockey skates are added to the equation, this challenge should drastically increase.

1.02 Postural Control in Ice Hockey Skates

Hockey skates, which in the simplest of forms consists of a boot, a blade holder, and a 3 mm wide blade, requires users to balance on the edge of a very thin piece of metal. Skating in ice hockey is a unique and challenging form of locomotion in which players are restricted to this very narrow BoS, compete within a dynamic environment, and are required to continuously manoeuver around oncoming opponents. On average, National Hockey League (NHL) players execute over 300 starts, stops, crossovers, sharp turns, or directional changes throughout the course of a game (Montgomery, Nobes, Pearsall, & Turcotte, 2004), movements that constantly shift their CoM in and outside of their BoS. Furthermore, during the basic task of forward skating, one must balance on only one skate blade for 80% of the stride (Pearsall, Turcotte, & Murphy, 2000), supporting themselves on a BoS that is roughly 34 times narrower than a barefoot BoS (Figure 1).

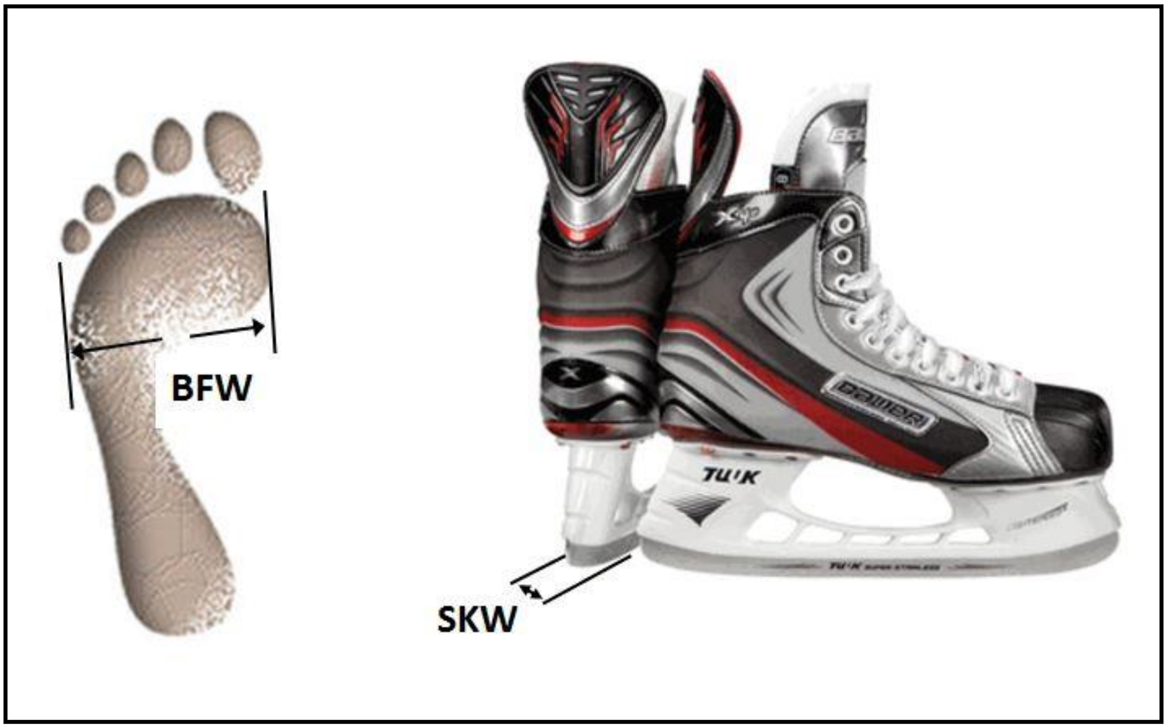

Figure 1: Representation of the average single leg base of support in skates and on bare feet. Skate blade width (SKW = 0.3 cm) and barefoot width (BFW = 10.3 cm) (Yu & Tu, 2009)

The skates themselves represent an inverted pendulum base; thus, one may consider standing in skates a compound inverted pendulum balance challenge (Figure 2). Clearly this is a difficult task and one that has yet to be explored.

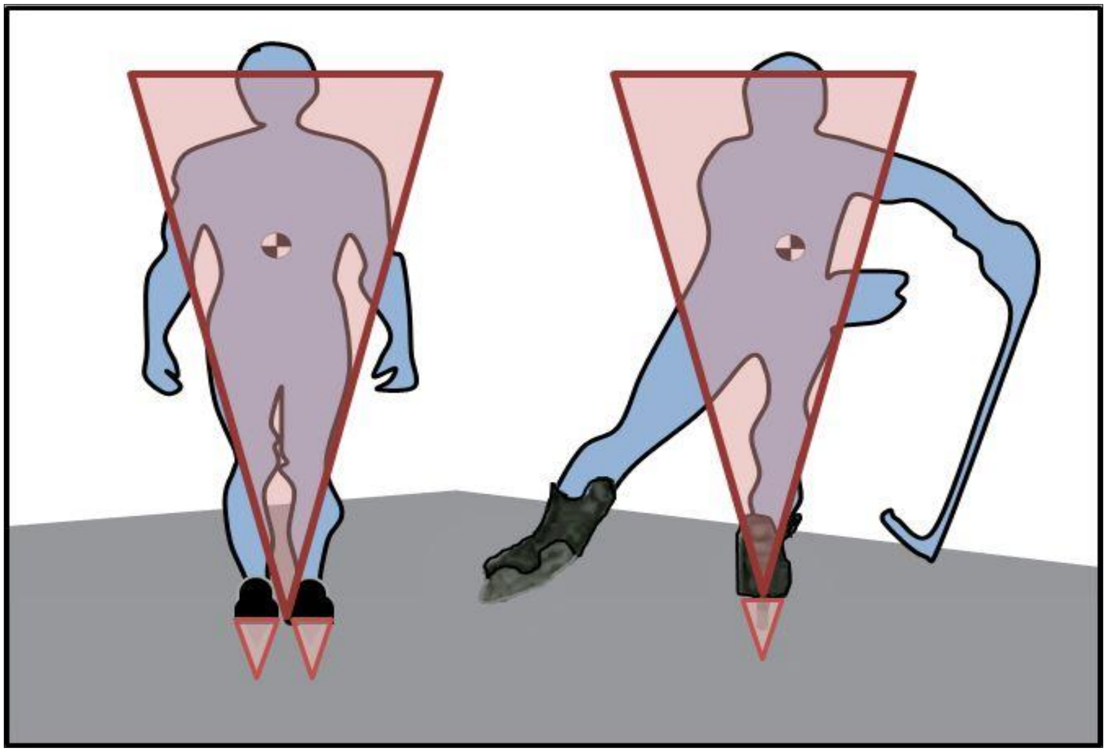

Figure 2: Double inverted pendulum model of postural control on ice hockey skates during double support (left) or single support (right) phases of the skating stride.

1.03 Literature on Postural Control in Ice Hockey/Skating

Few studies have touched on the subject of balance and postural control in ice hockey, and those that have, failed to directly assess postural control in skates. Bradley, Stotz, and Goodman (2003) found that hockey players utilized the same strategy of postural control for both large and small internal arm perturbations, whereas non players adopted different strategies based on the perturbation magnitude. Alpini, Hahn, and Riva (2008), who yielded some results in contrast to Bradley et al. (2003), studied hockey players with both eyes open and closed, while standing on either a form surface or a hard surface. Their most interesting finding was that in control and amateur subjects, the head was more stable than the trunk in all four conditions, however in elite subjects, the head was only more stable on the firm surface. This was thought to be because elite players adopt a different type of segmental postural control strategy on the destabilizing surface (foam), due to habituation to ice surface conditions. Figure skaters have also been studied, and it has been found that they perform better than controls on unstable surfaces (foam) but not any better on more stable surfaces, as they are accustomed to a challenging support surface due to their sport participation (Alpini, Mattei, Schlecht, & Kohen-Raz, 2008). Li, Xu, and Hoshizaki (2009) examined ankle proprioception in different athletic populations using a custom made device that rotated the foot into dorsiflexion, plantar flexion, inversion, or eversion. Subjects were in a seated position with their legs strapped into a thigh cuff suspension system that bore roughly 50 percent of their lower extremity weight. This was used in an attempt to eliminate any unwanted sensory cues from their feet contacting the surface of the plate. The surface where their feet were positioned then rotated in one of the four axes and subjects were instructed to press a hand trigger when they sensed any motion about their ankle and to identify the direction of that motion. Authors found that hockey players and ballet dances had significantly better proprioception in both dorsi/plantar flexion and inversion/eversion compared to runners and sedentary individuals. The reason for this difference was said to be because hockey and ballet both involve non-cyclic movements that are very challenging to the foot and ankle complex. Finally, in a study that used elder subjects, authors discovered that aged participants who were regular ice skaters showed postural control abilities similar to that of young non skaters (Lamoth & van Heuvelen, 2011). These few studies suggest that hockey players may adopt a different postural control strategy based on the testing surface or their skill level, and they may have superior proprioception and balance than other groups of individuals. However, there has yet to be a study that has examined postural control while wearing actual ice hockey skates, obviously an important contextual factor.

1.04 Footwear and Postural Control

The importance of incorporating hockey skates into the study of postural control in hockey players stems from studies pertaining to athletic footwear. Noé, Amarantini, and Paillard (2009) examined the effect that ski boots, a rigid type of footwear that mechanically restricts ankle motion, had on postural control strategies. They discovered that postural control was significantly improved (smaller CoP area) while wearing ski boots in static conditions, but no differences existed when they examined more dynamic conditions. Dynamic conditions challenged medio/lateral and antero/posterior stability using a wobble board device. This result was coupled with changes in the location of the COP and lead to altered muscle activation. Because both ski boots and hockey skates are stiff structures that limit ankle degrees of freedom, some similarities in the effects on postural control performance may be present. Even when regular footwear differences are compared, changes in postural control can still be seen. It is known that different running mechanics are displayed when comparing barefoot and shod running (Lieberman et al., 2010), that dynamic stability changes depending on whether people are tested in barefoot or while wearing footwear (Robbins, Waked, Gouw, & McClaran, 1994), and lastly, that different forms of everyday footwear have the ability to influence postural stability in both beneficial and detrimental ways (Menz & Lord, 1999). It is apparent that both static (no movement) and dynamic (surface translations/rotations) balance tasks are significantly affected by footwear, and therefore ice skates must be incorporated in order to truly understand the postural control mechanisms of hockey players. However, not only will the type of footwear affect an athlete's postural control, but so too will their level of athletic ability.

1.05 Postural Control of High and Low Calibre Athletes

There have been many cases where elite athletes have displayed superior balance or postural control over their recreational or less experienced counterparts. Although studies comparing elite and recreational level hockey players may be limited, research on other sports can shed light on the subject. Multiple studies of rifle shooters have shown that international or elite shooters possess better postural control than national or naïve shooters, as they show significantly less body sway in terms of CoP velocity (Konttinen, Lyytinen, & Era, 1999) and CoP amplitude (Era, Konttinen, Mehto, Saarela, & Lyytinen, 1996; Konttinen et al., 1999; Niinimaa & McAvoy, 1983). Elite soccer players display similar performance outcomes, showing both superior static and dynamic postural control in comparison to lower level amateur or regional players, as defined by smalled CoP area and velocities (Paillard & Noé, 2006; Paillard et al., 2006). Finally, highly proficient golfers also tend to exhibit greater static balance, as they display significantly smaller standard deviations of the ground reaction forces during unipedal stance (Sell, Tsai, Smoliga, Myers, & Lephart, 2007). Nevertheless, these examples of elite athletes demonstrating elite balance performance are not always the case. National and international alpine skiers have shown inferior balance in comparison to regional skiers when tested in static positions (Noé et al., 2009). The same result was also seen in surfers; when using static tests to assess postural control, there were no significant differences between elite and lower level surfers (Chapman, Needham, Allison, Lay, & Edwards, 2008; Paillard, Margnes, Portet, & Breucq, 2011). Interestingly though, when surface perturbations or balance board tests were introduced as a way to assess dynamic postural control, the opposite affect was demonstrated in skiers (Asaka et al., 2012) and surfers (Paillard et al., 2011) respectively, as the elite group showed better postural control than their less proficient counterparts. At the other end of the spectrum, in studies comparing varying levels of judoists, no significant differences existed between the elite and novice competitors (Paillard, Costes-Salon, Lafont, & Dupui, 2002). Evidently, the testing conditions, either static or dynamic, as well as how the postural control test relates to the context of the sport, plays a role in the relationship between high and low calibre athletes.

1.06 Rationale

It is common knowledge that maintaining stability while wearing skates is challenging, particularly to the novice skater; however, the specific postural control mechanisms that experienced skaters have developed is not well understood. Stability is fundamental to proficient execution of skating as well as relevant to player safety. There has yet to be a study that has examined postural control while wearing actual ice hockey skates, nor one that has measured CoM or CoP, standard assessment metrics of balance. Understanding how the body responds to instability created by blade-surface perturbations is essential to comprehending skating locomotion mechanics itself.

1.07 Objectives & Hypotheses

Therefore, the goal of this study is to (1) determine the body kinematic response, and the changes in the CoP and CoM as one attempts to regain balance after a surface perturbation; (2) to compare these strategies between bare foot (BF) and skate (SK) wearing conditions, and (3) to examine the differences, if any, between elite (ELT) and recreational (REC) level hockey players.

It is hypothesized that similar temporal patterns of body kinematics will be seen across groups and conditions; however, angular displacements will be exaggerated and therefore greatest in (A) recreational players and (B) during the skate condition. It is expected that elite hockey players will be better able to control their balance and upper body movements (i.e. trunk and hip angular displacement) through proper torque production of the lower limbs, following a perturbation.

The second hypothesis is that elite hockey players will display better balance both on skates and during barefoot conditions; however, the difference will be most pronounced during the skate conditions, due to elite players' greater repetitive and more intense experience in wearing skates. However, across both groups, it is expected that balance performance will be superior during barefoot conditions due to a lower CoM position and a more stable surface contact point.

1.08 Operational Definitions

Perturbation:	10cm horizontal displacement of the surface on which the subject is standing on.
Elite:	Players who are currently playing, or have played competitive junior, or university-level hockey.
Recreational:	Players who play in recreational leagues or drop-in hockey (McGill intramurals or pickup hockey players).
Resting CoP/CoM:	Center of pressure or center of mass during the one second window prior to perturbation onset.
Perturbation CoP/CoM:	Center of pressure or center of mass position during the perturbation and one second following the end of the perturbation.

1.09 Limitations

- Ankle and foot markers were placed on the skate boot in order to be visible, therefore it was not an exact 3D representation of the ankle complex
- Only horizontal surface perturbations were studied
- Testing was done on a low friction polyethylene surface, not real ice.

1.10 Delimitations

- Only male subjects were used in this study
- Only subjects between the ages of 18 and 30 were recruited.
- Only four perturbation directions were tested.
- Subjects heel width was kept constant at 15% of their shoulder width
- Subjects were not be wearing their own hockey skates, but used ones provided to them.

1.11 Ethical Considerations

The potential risks of this research were minimal. Subjects were instructed to simply maintain their balance during a perturbation of their support surface. The velocity and acceleration that the platform translated was not large enough to elicit a fall or even a stumble, as we used similar perturbation magnitudes as previous studies (Asaka et al., 2012; Hilderley, 2011; Trivedi, 2010). Still, as a precaution all subjects wore a safety harness to prevent fall injuries. Calculating body kinematics included taping adhesive markers to the skin; however this was not invasive and was an extremely common technique used for quantifying human movement.

All the personal information collected during the study was encoded using a numerical coding system in order to keep subjects' confidentiality. These records will be maintained at the Biomechanics Laboratory by Dr. David Pearsall for five years after the completion of the project, and will be destroyed afterwards. Only members of the research team (principal researcher, faculty supervisor, and lab technician) will have access to them. For presentation and publication purposes, subjects' identities will remain anonymous. Data will be stored on the computer for further analysis and will be kept confidential by the experimenter and lab technician. Further storage will be on a mobile storage unit and securely stored in a locked cabinet. An Information and Informed Consent form will be required from each subject prior to any data collection.

1.12 Contribution to the field

The current research will provide better understanding of how hockey players are able to maintain balance on a narrow skate blade on a low friction surface. No study to date has looked at postural control while actually wearing ice hockey skates; therefore, this is a very novel approach. Through the comparison of elite and recreational level players, we should be able to determine whether different calibre players utilize similar or contrasting strategies to maintain upright balance, and whether one strategy is superior to another. By comparing subjects when barefoot and in skates, we can determine if it is in fact more difficult to remain balanced in hockey skates, and examine the kinematic or kinetic changes to adjust to the different footwear conditions. These outcomes can potentially lead to optimizing balance strategies for athletes, or lead to helping elite athletes analyze deficiencies in their postural control.

Chapter 2: Methods

2.1 Subjects

A sample of elite (N =11) and recreational (N = 13) level hockey players ranging in age form 20-29 years, volunteered to participate in this study (Table 1). All participants were screened prior to the study to ensure no previous lower limb injury or medical condition would put them at risk (Appendix A). Subjects who have a previous history of neuromuscular disorders, balance disorders or sensory loss were excluded. Subjects who have had a significant knee or ankle injury that has prevented them from playing within the past year were excluded.

Significant differences were found in playing experience between elite (19.3 +/- 4.1 years) and recreational level (14.2 +/- 3.5) participants (*p = 0.003).* Informed consent was obtained from all participating subjects and ethics was approved prior to their involvement in the study.

Table 1: Participant characteristics

Variable	Age (years)	Height (m)	Mass (kg)	Experience (years)
		Elite calibre (*n* = 11)		
Mean	24.7	1.80	83.0	19.3
SD	2.6	0.04	7.7	4.0
		Recreational Calibre (*n* = 13)		
Mean	23.0	1.82	77.2	14.2
SD	2.3	0.08	10.5	3.5

2.2 Equipment

All experiments were conducted in the Balance and Voluntary Movement Laboratory at McGill University. To allow for optimal lighting conditions, blinds were constantly drawn and areas of the room were covered in dark cloth to reduce unwanted reflections.

A seven camera MX3 Vicon Nexus™ 1.7.2 motion-capture system (Vicon Motion Systems, Oxford, UK) collecting at 200 Hz was used in order to capture full body kinematic data from each subject. Two tri-axial force plates (model FP4060, Bertec, Columbus OH) recorded ground reaction forces from each foot in the mediolateral (x, M-L), anteroposterior (y, A-P), and vertical (z) axes at a sampling rate of 1000 Hz. Each force plate surface was covered with a 4mm thick polyethylene artificial ice surface, to allow for skates to be worn. The reported coefficient of friction of the artificial ice surface is 0.27 ("Overview," 2009). Both force plates were embedded within a moveable platform, capable of producing perturbations in multiple directions in the horizontal plane. Padded mats were placed around the two force plates to protect the surface of the platform from scratches (Figure 3) and more important, protect the subject from slipping and falling if a recovery step had to be taken. An A-C servo motor controlled the platform position, velocity, and acceleration, while an external trigger synchronized all three systems, both using custom scripts written in NextMove ESB Workbench (Baldor Electric Co., Fort Smith, AR) and Labview (National Instruments, Austin, TX).

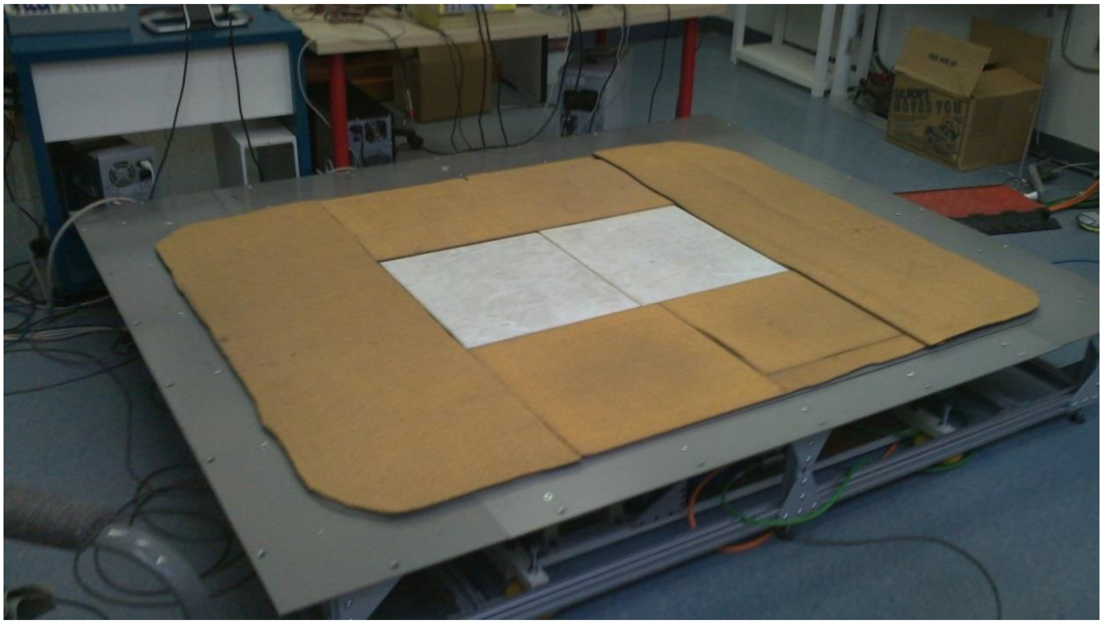

Figure 3: Perturbation platform with two embedded force plates (center). Each force plate has a polyethylene artificial ice surface adhered to the surface and is surrounded by protective rubber mats.

2.3 Experimental Protocol

After reading and signing the informed consent document, the subject was given tight fitting spandex clothing to wear throughout the testing session. This allowed for optimal adhesive conditions for the reflective markers. Anthropometric measurements were then collected from each subject (Appendix B). Following this, 39 reflective markers were affixed to various landmarks on their body in accordance with Vicon's PluginGait model. Subjects were then strapped into a retractable safety harness that was connected to the ceiling, directly above the force platform.   Although perturbations were not large enough to induce a fall, the harness ensured subject safety in any case. Finally, subjects were given a hockey stick for contextual realism; however, they were not allowed to rest it on any part of their body, or place it on the platform for support. Subjects' heel width was standardized to 1.15 times the distance of the shoulders (i.e. between the left and right acromioclavicular joints). Initial body position was a hybrid of a completely upright posture and a full-squat hockey posture (Figure 4). Instructions were to get in a ready position, with their knees and hips slightly bent, their chest forward, their head up, their stick comfortably in front of or across their body, and their eyes straight ahead focusing on a visual target positioned at eye level on the wall. Subjects were also shown an image (Figure 4) of the position to adopt. This position was chosen to ensure some form of standardization, however allowing for slight individual differences based on personal preference.

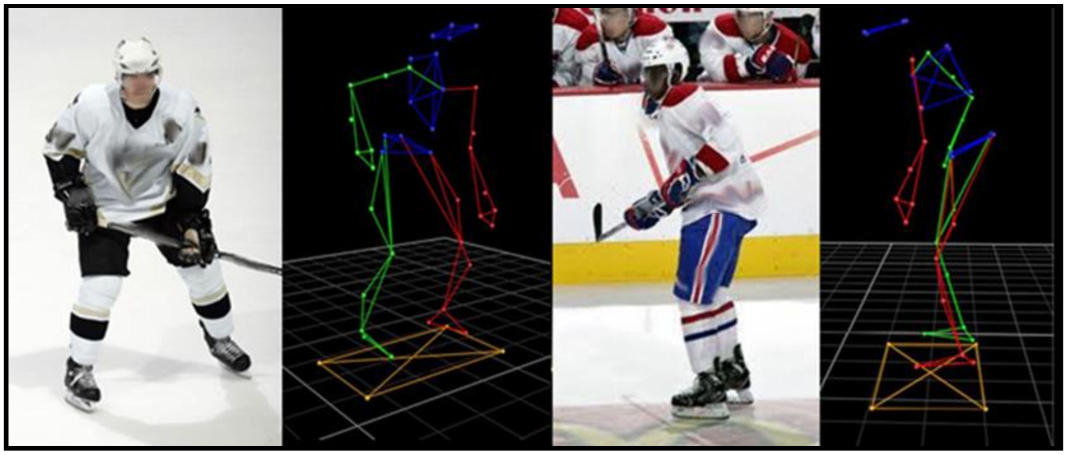

Figure 4: Initial body posture subjects were both verbally and visually instructed to position themselves in.

Prior to actual data collection, subjects were given a familiarization period in which the velocity and acceleration of the moving platform was increased incrementally until the real testing conditions were reached, the subject felt relatively comfortable, and their responses were consistent. This time period was no longer than ten minutes and was consistent with past research (Asaka et al., 2012; Leonard, 2009, 2011; Mansfield & Maki, 2009; Runge, Shupert, Horak, & Zajac, 1999). During the actual experimental conditions, the platform displacement (10cm), velocity (40cm/s), and acceleration (392cm/s^2) was kept constant. These values were similar to what has been used in previous research to elicit quantifiable postural responses without making subjects fall or forcing them to take a step (Asaka et al., 2012; Hilderley, 2011; Trivedi, 2010). See Figure 5 for subject setup.

Figure 5: Anterior (left) and posterior (right) views of subject positioned on the perturbation platform.

During condition one, subjects stood on the artificial ice surface wearing hockey skates provided to them, while during condition two, they stood barefoot on the same surface. The order was randomized for each subject but all other protocol was identical.

During phase one of the experiment, static postural control was assessed. Subjects were instructed to get in the ready position and then two trials of 30 seconds each were collected, with the subjects attempting to stay as still and balanced as possible.

During phase two, dynamic postural control was assessed. During these trials, the subjects began in the same initial position, but the surface translated beneath their feet and they had to attempt to regain their balance. A total of four translation directions were used, with 90° separating each direction (Figure 6). Each translation was randomly executed and repeated four times, for a total of 16 dynamic trials per footwear condition. Data was collected for one second prior to the translation onset (baseline) and continued for three seconds after the translation onset, for a total trial time of four seconds. This ensured all balance recovery movements would be captured. If the subject took a recovery step following a perturbation, the trial was repeated at the end of the respective phase.

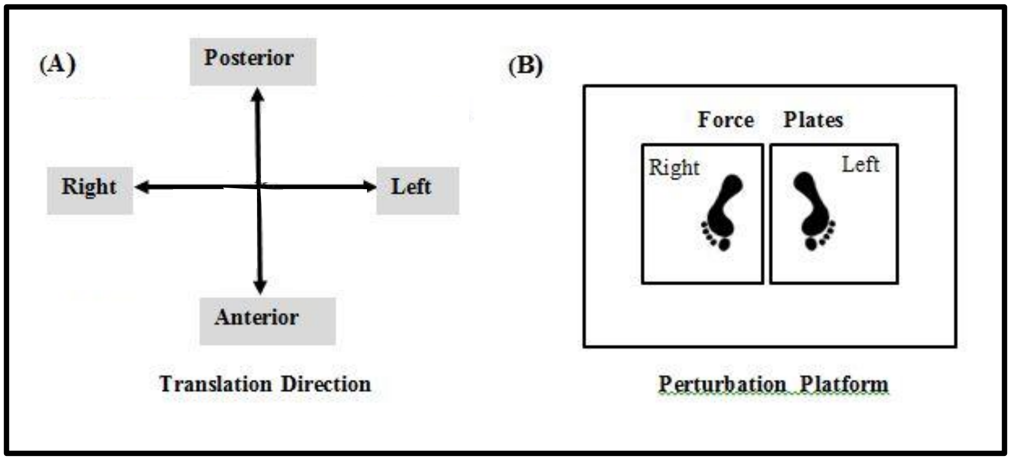

Figure 6: Translation directions of the perturbation platform (A) and subject positioning on the force plates (B).

2.4 Research Design

This study used multiple one-way ANOVA's. The independent variables included footwear (SK / BF) and skill level (ELT / REC). See Table 2. The dependant variables were CoP delta, CoP excursion, CoM delta, and multiple kinematics angles. See Table 3 for a complete description of all the dependent variables.

Table 2: Independent variables with associated levels

Independent variables	Levels
Calibre	Elite
	Recreational
Footwear	Hockey skates
	Barefoot

Table 3: Description and calculation of dependent variables

Variable	Variable description	Value
CoP delta[A] (CoPx,y)	Max displacement (X[B] and Y[C]) from resting CoP to perturbation CoP	Mean ± SD,
CoP excursion (CoPex)	Total path length of CoP from resting CoP to perturbation CoP	Mean ± SD
CoM sway[A] (CoMx,y,z)	Max displacement (X[B],Y[C], and Z[D]) from resting CoM to perturbation CoM	Mean ± SD
Ipsi Ankle Angle	Maximum change in angular displacement between foot and shank (sagittal and frontal plane)	Mean ± SD
Ipsi Knee Angle	Maximum change in angular displacement between the shank and the axis extended from the thigh (sagittal and frontal plane)	Mean ± SD
Ipsi Hip Angle	Maximum change in angular displacement between the thigh and the horizontal axis projected from the pelvis (sagittal and frontal plane)	Mean ± SD
Ipsi Thorax Angle	Maximum change in angular displacement between trunk and vertical global axis (sagittal and frontal plane)	Mean ± SD
Ipsi Shoulder Angle	Maximum change in angular displacement between upper arm and thorax segment (sagittal & frontal plane)	Mean ± SD

[A] Standard Deviation of CoP and CoM used during Static trials

[B] Delta X used during Ipsi/Contra perturbations

[C] Delta Y used during Anterior/Posterior perturbations

[C] Delta Z used during all perturbations. Represents the CoM height as % of subjects' height

2.5 Data Acquisition

LabVIEW™ Version 10.0 (National Instruments®, Austin, Texas) software was used to control the perturbation platform, and also to activate the external synchronization trigger. Vicon Nexus™ 1.7.2 was used to collect and reconstruct the Vicon MX camera data, and to collect data from both force plates. MATLAB™ (7.10.0, R2010a, MathWorks, Inc., Massachusetts, U.S.A.) software was used to post-process the data, including re-sampling, filtering, and calculating all the dependent variable.

2.6 Data Processing and Analysis

Following data collection, the raw marker coordinates and force plate data were imported into Vicon Nexus™ 1.7.2 for post processing. The raw trajectories of each marker were reconstructed in order to create a 3D model. This data was then transferred to Vicon IQ 2.5 for labelling and gap filling. Gaps, which were defined as sections of trajectories that were missing coordinates, were filled using either virtual points or spline functions. If the gap occurred on a rigid structure of four or more markers (i.e. head, hips, or trunk) a virtual point was created. The virtual point function uses the coordinates from three other markers, which are stationary relative to the missing marker, in order to calculate the position of the fourth marker. If the gap didn't occur on a rigid structure with more than four markers (i.e. arms, legs, hands, or feet) the spline function was used. This function employs a set of polynomial equations which uses the trajectory of the markers both before and after the gap in question in order to create a smooth and appropriate fit. If neither the virtual point, nor the spline function could accurately fill the gap, the trial was deemed unfit, and was deleted. Following this processing and labelling, the data was imported into MATLAB™ (7.10.0, R2010a, MathWorks, Inc., Massachusetts, U.S.A.) for further calculations.

Force Plate

Force plate analog channels were down-sampled to 200Hz to match the Vicon motion capture data. Force plate channels (F_x, F_y, F_z, M_z, M_y, M_z) were then filtered using a 4th order low-pass Butterworth filter with a cut-off frequency of 10 Hz. Cut off frequencies of the filter were chosen based on the Fast Fourier transformation of the raw analog data. The local reference frame of each force plate was then adjusted to correspond to the global reference frame that the Vicon system was using, to ensure that all motion and applied forces in each direction would have an output along the same axis. The center of pressure in the x and y directions was then calculated for each force plate. The center pressure equations for one force plate are presented below

[1] $$CoPx1 = \frac{-h \times (Fx1 - My1)}{Fz1}$$

[2] $$CoPy1 = \frac{-h \times (Fy1 + Mx1)}{Fz1}$$

where *h* represents the thickness of the polyethylene surface attached to the force plate. The net center of pressure in both the x and y direction was then calculated using the formula from Winter et al. (1996) which combines the input from force plate one and two.

[3] $$CoPnetx = CoPx1 \times \left(\frac{Fz1}{Fz1 + Fz2} \right) + CoPx2 \times \left(\frac{Fz2}{Fz1 + Fz2} \right)$$

[4] $$CoPnety = CoPy1 \times \left(\frac{Fz1}{Fz1 + Fz2} \right) + CoPy2 \times \left(\frac{Fz2}{Fz1 + Fz2} \right)$$

Equations 3 and 4 were used for all calculations pertaining to center of pressure variables. Center of pressure location was calculated from the origin (0,0) located at the center of each force plate, without respect to foot placement. See Table 3 for a description of the center or pressure variables.

Kinematics

Marker data was filtered using a 4th order low-pass Butterworth filter with a cut-off frequency of 14 Hz, in order to eliminate any unwanted noise. The center of mass position of the subject during each trial was calculated in Nexus™ using the full body plug-in-gait model. This represented a weighted sum of each segments' center of mass. However, due to the 10cm displacement of the platform during the dynamics trials, the CoM displacement was being overestimated, as it was calculated relative to the global reference frame and not the local reference frame. To account for this, CoM position relative to the right anterior force plate marker (RA) was used to calculate the net CoM displacement, as this marker translates with the force plate. Joint angles were calculated in both the sagittal and frontal planes in order to determine the postural control strategies that players adopted. This included the trunk segment (thorax), shoulder, hip, knee, and ankle angle (Figure 7). The respective values were obtained from Vicon's™ Plug-in-Gait model.

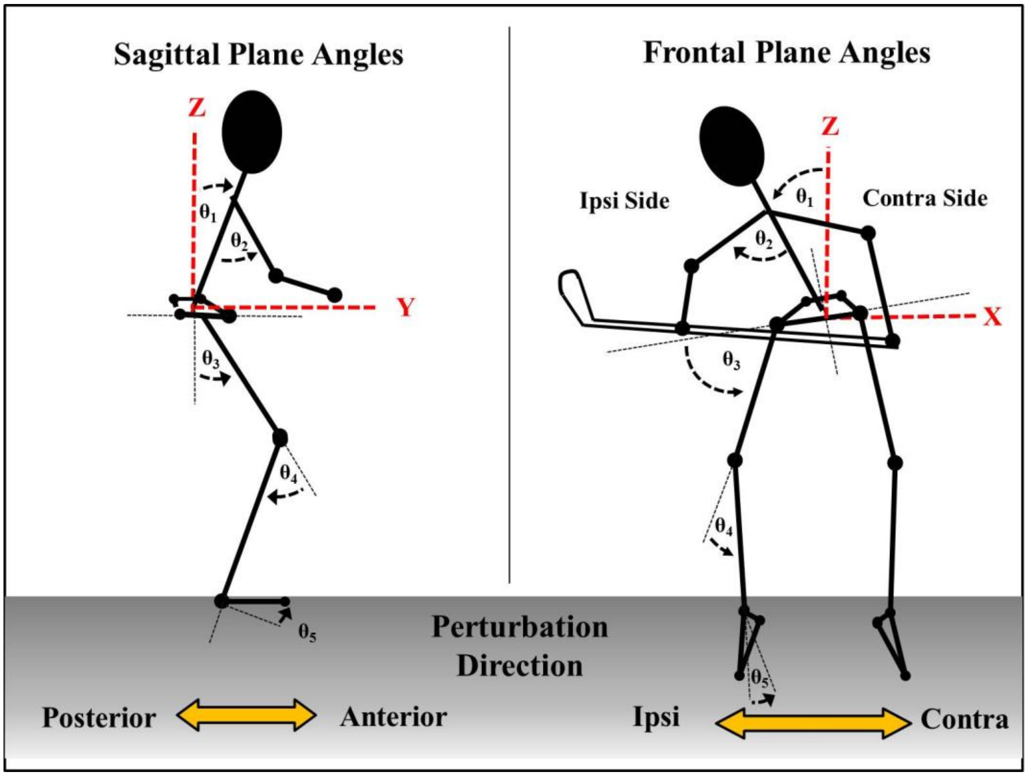

Figure 7: Joint angle definitions in the sagittal (left) and frontal (right) plane for a right handed shooter. Note: Θ_1 = Thorax, Θ_2 = Shoulder, Θ_3 = Hip, Θ_4 = Knee, Θ_5 = Ankle.

Due to the participation of both left and right shooters, coupled with the perturbations occurring in both the left and right direction, kinematics and perturbation direction were normalized to the shooting side, using the Ipsi / Contra nomenclature. For example, if the subject was a right handed shooter (right hand holds stick's mid-shaft), a perturbation TOWARDS the right was considered an Ipsi Perturbation, while angles on the right side of their body were labelled as Ipsi Angles. Alternatively, a perturbation TOWARDS the left would result in a Contra Perturbation, and Contra Angles respectively. See Figure 7.

2.7 Statistical Analysis

This study used multiple one-way ANOVA's. The independent variables included footwear (SK / BF) and skill level (ELT / REC). Individual ANOVAs were conducted on the seven kinematic dependant variables as well as on the two kinetic dependant variables for each for the five perturbation conditions. SPSS (IBM® SPSS® Statistics, Version 21.0) was used to conduct all analyses.

Chapter 3: Results

The current study aimed to quantify and determine whether postural control differences existed between two different footwear conditions, and between two different player calibre conditions.

3.1 Static Condition

Players in skates tended to exhibit greater variability in their center of mass sway (CoMx SD) than those in bare feet, $F(1,47) = 4.13$, $p = .048$; however, barefoot conditions yielded greater variability in the center of pressure (CoPx SD) location $F(1,47) = 4.42$, $p = .041$ (Table 4). Variability was defined by the standard deviation. There were no significant differences across calibre conditions, nor was there an interaction effect. See Figure 8 for a comparison of significant differences. Although these differences were statistically significant, the differences were in the range of ~1 mm, therefore can be considered irrelevant.

Table 4: Means and standard deviations of the five dependent variables during the static condition

		Center of Pressure (cm)			Center of Mass (cm)	
		CoPx SD	CoPy SD	CoPex	CoMx SD	CoMy SD
Elite	Mean	0.07	0.61	35.1	0.25	0.59
	SD	0.04	0.26	16.3	0.11	0.24
Rec	Mean	0.07	0.52	38.5	0.27	0.53
	SD	0.04	0.22	34.2	0.27	0.24
BF	Mean	**0.08***	0.60	39.2	**0.20***	0.57
	SD	0.05	0.27	38.9	0.08	0.27
SK	Mean	**0.06***	0.52	34.9	**0.33***	0.53
	SD	0.03	0.21	08.6	0.30	0.21

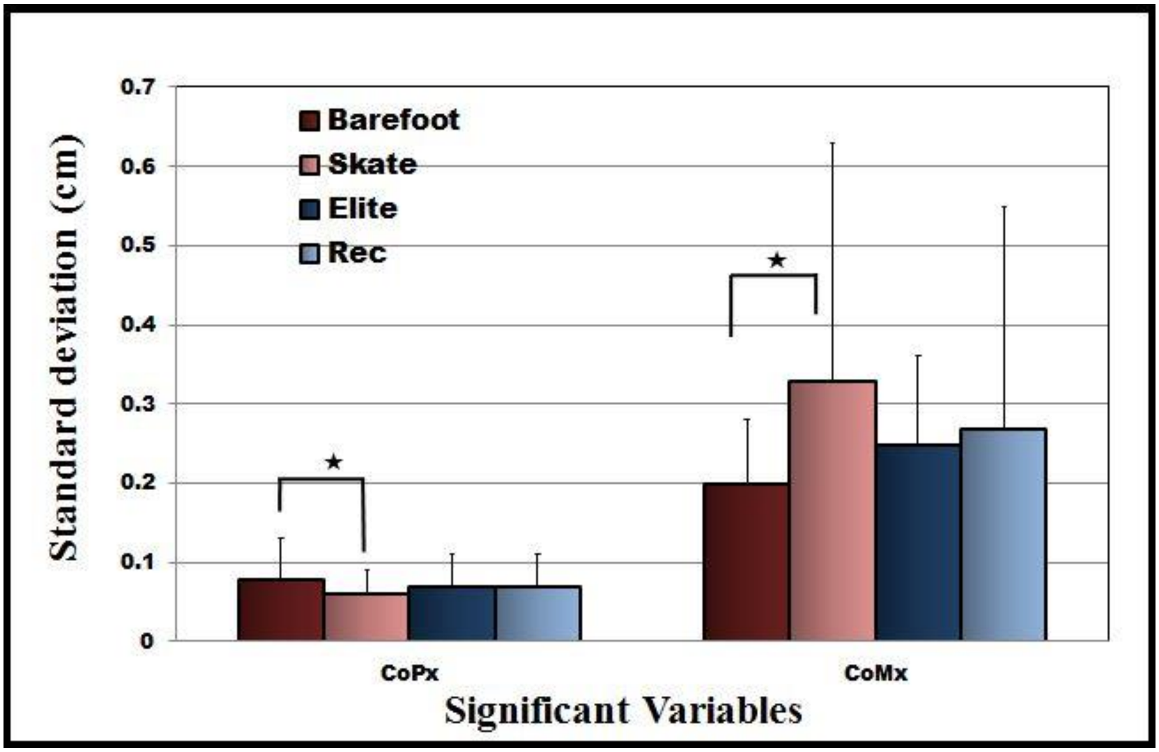

Figure 8: Magnitude of the average center of pressure standard deviation and the center of mass standard deviation during the Static condition (+SD bars; * $p<0.05$).

3.2 Ipsi Perturbation

Perturbation Profiles

The frontal plane kinematic and center of mass profiles were quite similar across calibres and footwear conditions (Figure 9). Delta values of the joint angles and CoPx, as well as the sway values of CoMx and CoMz, were defined as the greatest change in displacement (max – min) over the course of the perturbation (300ms) as well as the recovery period afterwards (600ms). Center of pressure excursion values were calculated over the same time period, however they represent the total path length of the CoPx and CoPy trace.

Immediately following the Ipsi perturbation, the lower limbs started to move due to the translation of the force plate. The ipsi ankle complex began to evert minimally (i.e. less than 2°) followed by knee and hip abduction (less than°). The CoMx shifted up to 7.5 cm contra-laterally (i.e. opposite to the direction of the perturbation). The CoMz remained stable. Twenty frames (100ms) after the onset of the perturbation, the thorax tilted 4° in the opposite (contralateral) direction corresponding with 4° of ipsi shoulder abduction. Approximately 25 frames (125ms) after the start of the perturbation, an opposite movement at the hip, knee, and ankle joint was produced. The CoMx began to sway back towards the center of the stance. The thorax and shoulder were the last to respond to the perturbation and bring themselves back to their original position.

In terms of center of pressure, at the onset of the perturbation, CoPx shifted 1 to 2 cm opposite to the perturbation. This corresponded to the CoMx movement, as the center of mass also swayed towards the direction opposite of the perturbation. Roughly 20 fames (100ms) before the force plate stopped translating, the center of pressure began to travel back towards the middle of the stance, in an attempt to bring the body back to its original state.

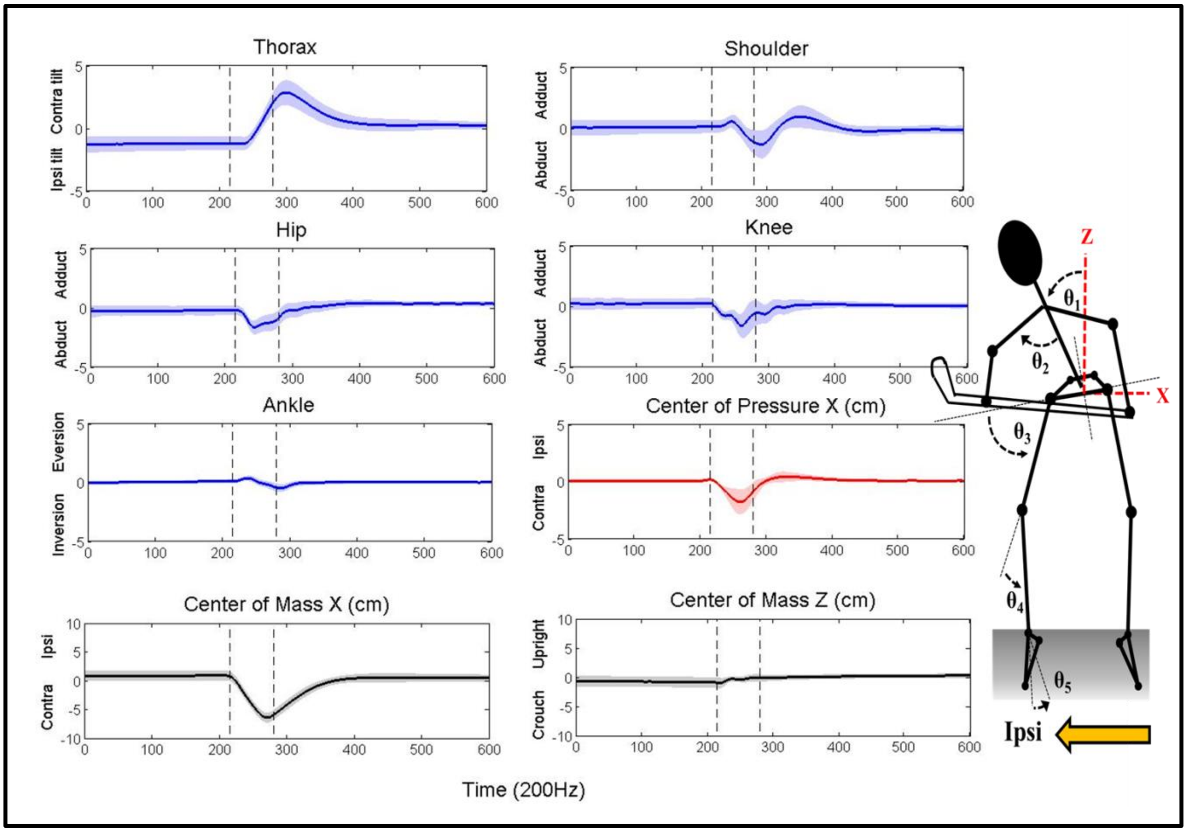

Figure 9: Average kinematic, center or mass, and center of pressure profiles (mean +/- SD) of all subjects during the Ipsi perturbation. Vertical lines indicate start and end of perturbation. Note: Initial start position was normalized.

Statistical results

There was a significant difference in the CoMx, $F(1,47) = 49.95$, $p = .000$ between barefoot and skate conditions, as players in skates tended to exhibit greater center of mass sway than those in bare feet.

Similar to the kinematics, analysis of the center of pressure variables resulted in a statistically significant difference across footwear conditions for CoPx, $F(1,47) = 6.85$, $p = .012$, as barefoot conditions yielded greater center of pressure displacement values in comparison to skate conditions. Calibre conditions were also significantly different, CoPx, $F(1,47) = 4.98$, $p = .031$, with elite players showing greater CoPx values than recreational players. Although significant differences were found in both CoMx and CoPx, note the small magnitude (~1cm). See Table 5 for means and standard deviations. See Figure 10 for comparison of the significant differences.

Table 5: Posture displacements (means and standard deviations) of the nine dependent variables during the Ipsi perturbation.

Factor		Center of Pressure (cm)		Frontal Kinematics (θ°)					Center of Mass (cm)	
		ΔCoPx	CoPex	Thrx	Shldr	Hip	Knee	Ank	ΔCoMx	ΔCoMz
Elite	Mean	**-2.3***	11.3	4.6	-3.0	-2.8	-3.1	-1.2	-7.2	0.2
	SD	1.2	03.6	1.7	1.2	1.1	1.1	0.4	0.7	0.8
Rec	Mean	**-1.5***	10.4	4.1	-3.8	-3.0	-3.2	-1.1	-7.4	0.0
	SD	1.4	03.0	1.2	1.1	0.8	0.7	0.6	0.6	0.7
BF	Mean	**-2.3***	11.3	4.1	-3.7	-3.0	-3.3	-1.0	**-6.8***	0.1
	SD	1.3	03.9	1.4	1.3	1.0	1.1	0.4	0.4	0.8
SK	Mean	**-1.4***	10.3	4.6	-3.1	-2.8	-3.1	-1.1	**-7.7***	0.2
	SD	1.3	02.5	1.5	1.1	0.9	0.7	0.6	0.5	0.7

*p<0.05, See Figure 9 for indication of (-) direction

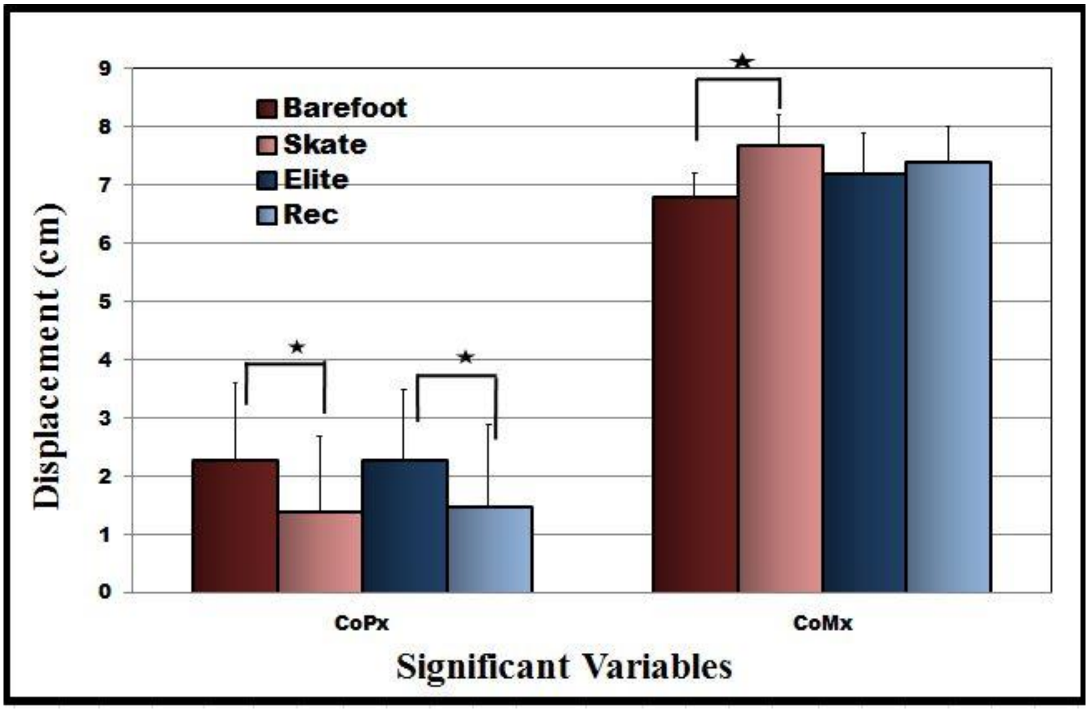

Figure 10: Magnitude of the average center of pressure delta and center of mass sway values during the Ipsi perturbation (+SD bars; * $p<0.05$).

To summarize the responses due to the Ipsi perturbation, no differences in the kinematic variables were observed; however, greater changes in CoMx sway were seen in skates compared to barefoot (7.7 vs. 6.8cm). Conversely, greater changes in CoPx were seen in bare feet compared to in skates (2.3 vs. 1.5 cm), while elite players showed greater changes in CoPx compared to recreational players (2.3 vs1.5 cm).

3.3 Contra Perturbation

Perturbation Profiles

The frontal plane kinematic and center of mass profiles displayed similar trends across calibres and footwear conditions (Figure 11).The Contra perturbation was in the opposite direction of the Ipsi perturbation, and therefore a reciprocal pattern of joint kinematics and center of mass movement was seen. Immediately following the perturbation, the lower limbs started to move due to the translation of the force plate. The ankle complex began to invert minimally (< 2º) followed by hip and knee adduction (< 3º) . At the same time the CoMx swayed laterally by roughly 7cm, opposite to the direction of the perturbation. Once again, the CoMz remained relatively stable. Approximately 20 frames (100ms) after the onset of the perturbation, the thorax tilted 4º in the opposite direction while the shoulder adducted by roughly 3º. Approximately 25 frames (125ms) after the onset of the perturbation, an opposite movement at the hip, knee, and ankle joint was produced. Following this, the CoMx began to sway back towards the center of the stance, followed by the thorax and shoulder, as they were the last to respond to the perturbation.

The center of pressure profiles also displayed similar trends across calibres and footwear conditions (Figure 11). At the onset of the perturbation, CoPx shifted 1 to 2 cm in the direction opposite to that of the perturbation. Similar to the Ipsi perturbation, this corresponded to the center of mass position, as the CoMx also swayed laterally, opposite to that of the perturbation. Approximately 20 frames (100ms) prior to the end of the perturbation, the center of pressure shifted back towards the middle of the stance, in an attempt to bring the body back to its original state

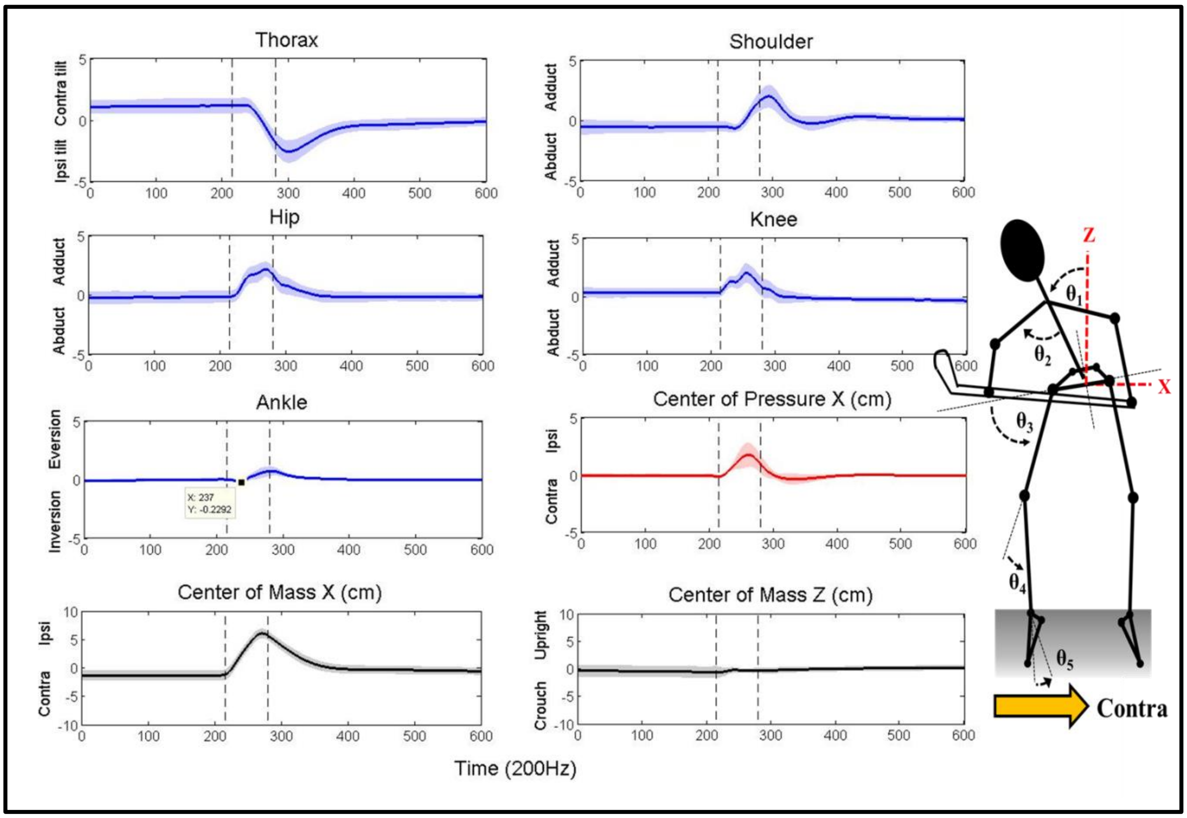

Figure 11: Average kinematic, center or mass, and center of pressure profiles (mean +/- SD) of all subjects during the Contra perturbation. Vertical lines indicate start and end of the perturbation. Note: Initial start position was normalized.

Statistical Results

A series of one-way ANOVA's on each of the nine dependent variables was conducted. Across footwear conditions, there was a significant difference in CoMx, $F(1,47) = 61.39$, $p = .000$, with skate conditions yielding greater center of mass sway. There was also a significant difference in the ankle angle, $F(1,47) = 7.26$, $p = .010$ between barefoot and skate conditions, as subjects experienced greater change in ankle eversion while in skates. However, the magnitude of this difference could be seen as negligible.

In terms of center of pressure, there was a significant difference across calibre for CoPx, $F(1,47) = 8.67$, $p = .005$, with elite players displaying greater center of pressure values. Across footwear conditions there was also a significant difference in CoPx, $F(1,47) = 11.54$, $p = .001$, as barefoot conditions also yielded greater center of pressure values (Table 6). In terms of the CoPex, no main effect for either calibre or footwear was found, but there was a significant interaction. Upon further analysis of the interaction, it was determined that elite players on bare feet showed significantly greater CoPex (12.9 cm) than both elite players on skates (9.7 cm) and recreational players on bare feet (9.7cm). See Figure 12 for a comparison of the significant differences.

Table 6: Posture displacements (means and standard deviations) of the nine dependent variables during the Contra perturbation.

Factor		Center of Pressure (cm)		Frontal Kinematics (θ°)					Center of Mass (cm)	
		ΔCoPx	CoPex	Thrx	Shldr	Hip	Knee	Ank	ΔCoMx	ΔCoMz
Elite	Mean	**2.4***	11.0	-4.2	3.1	3.6	3.3	1.3	7.4	0.3
	SD	1.3	03.3	1.5	1.4	0.8	1.1	0.7	0.7	0.6
Rec	Mean	**1.4***	10.2	-3.8	3.6	3.5	2.9	1.1	7.4	0.1
	SD	1.4	03.5	0.9	1.0	1.0	0.8	0.5	0.6	0.7
BF	Mean	**2.5***	11.1	-3.9	3.6	3.3	2.9	**0.9***	**6.9***	0.2
	SD	1.3	03.4	1.0	1.1	1.0	1.0	0.4	0.3	0.7
SK	Mean	**1.3***	10.3	-4.1	3.2	3.8	3.3	**1.4***	**7.8***	0.2
	SD	1.4	02.9	1.5	1.3	0.8	1.0	0.7	0.5	0.7

* $p < 0.05$, See Figure 11 for indication of (-) direction

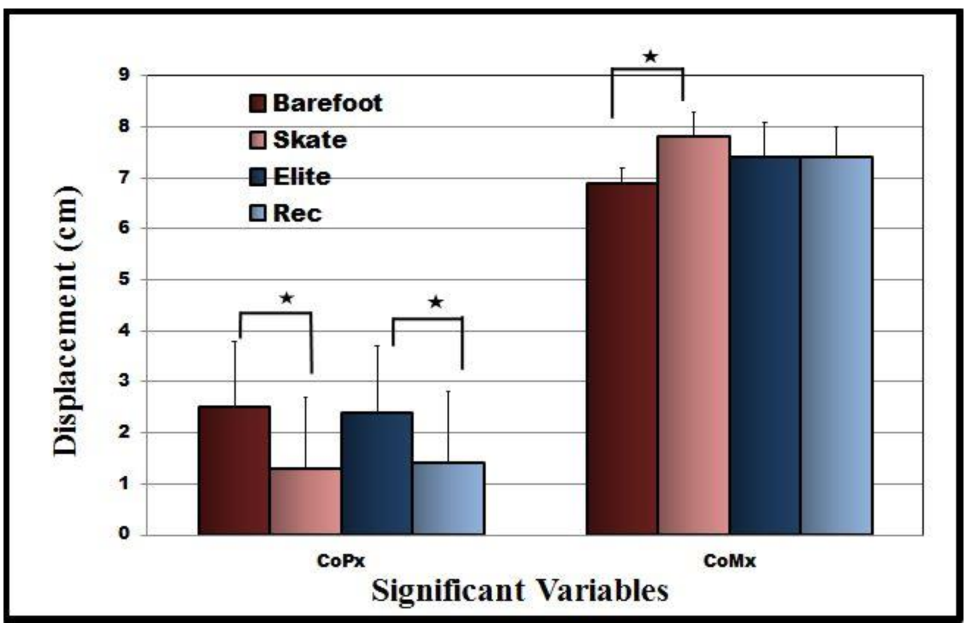

Figure 12: Magnitude of the average center of pressure delta and center of mass sway values during the Contra perturbation (+SD bars; p<0.05).

To summarize the responses to the Contra perturbation, no differences in the kinematic variables were observed; however, greater changes in CoMx were observed in skates compared to barefoot (7.8 vs 6.9 cm). Conversely, greater changes in CoPx were seen while barefoot compared to in skates (2.5 vs. 1.3 cm), while elite players showed greater changes in CoPx compared to recreational players (2.4 vs1.4 cm). An interaction was also observed which followed the same trend of CoP_{BF} > CoP_{SK} and the trend of CoP_{ELT} > $CoP_{REC,}$ as elite players on bare feet showed significantly greater CoPex (12.9 cm) compared to both elite players on skates (9.7 cm) and recreational players on bare feet (9.7cm).

In terms of the lateral perturbations (Ipsi-Contra), the variables showed similar results. By comparing the Ipsi perturbation graph with the Contra perturbation graph you can see that both perturbations showed almost identical trends in CoPx and CoMx, with a few significant differences observed (Figure 13).

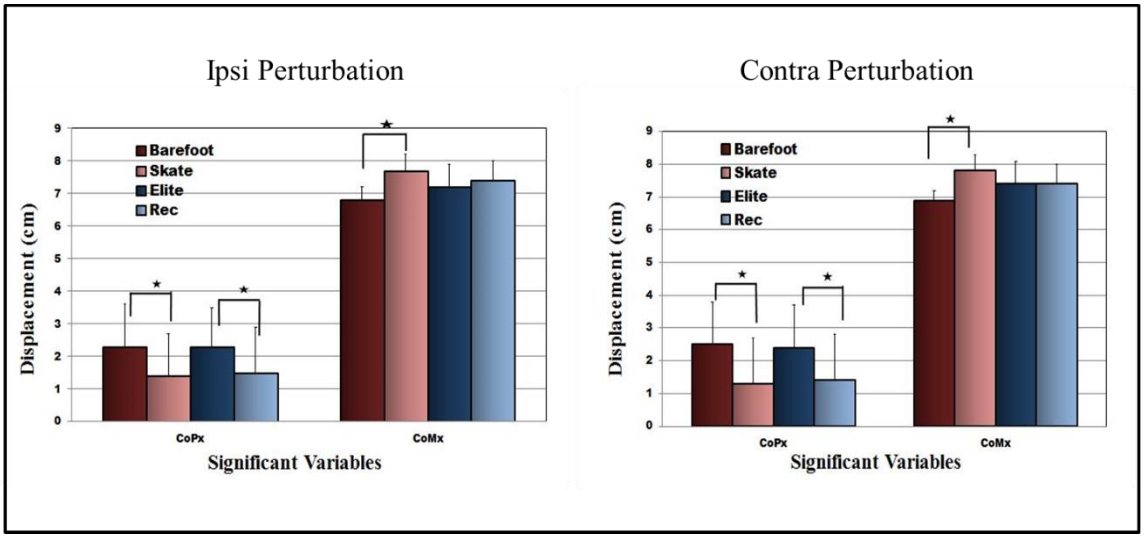

Figure 13: Comparison of the main effects in the Ipsi and Contra perturbation (+SD bars; $p<0.05$).

3.4 Posterior Perturbation

Perturbation Profiles

The sagittal plane kinematic, center of mass, and center of pressure profiles were similar across calibres and footwear conditions (Figure 14).

Immediately following the Posterior perturbation, the ankle complex dorsiflexed by approximately 3º, the knee flexed minimally (< 1º), and the hip extended (2º), all passively due to the force plate translation. During this time, the thorax, shoulders and CoMz were relatively stationary, however the CoMy swayed forward by approximately 8.5 cm, opposite to the direction of the perturbation. Twenty frames (100ms) after the onset of the perturbation, the ankle began to plantar flex, undergoing a 4º change, essentially creating a counter torque in an attempt to stop the body from continuing to rotate anteriorly. The knee and hip then started to respond, undergoing roughly 5º of extension and flexion respectively. The thorax, which responded after the hip, knee and ankle, tilted 9º anteriorly, while the shoulder flexed by 7º, moving the arms anteriorly. In comparison to the Ipsi and Contra perturbations, none of the limbs responded by moving back towards their initial position until *after* the perturbation ceased. Once the platform stopped translating, the opposite movement occurred at each limb in order to bring the body back to its starting position. The ankles plantar flexed, the knees flexed, while the hips extended. Simultaneously, the thorax and shoulder both extended, while the CoMy swayed back towards its initial position.

In terms of center of pressure, immediately following the onset of the Posterior perturbation, the CoPy shifted towards the front of the feet, corresponding with the anterior sway of CoMy. The center of pressure movement continued until approximately 10 frames (50ms) before the force plate stopped translating, at which time the subject began to recover and shift the center of pressure back towards its initial position. The total anterior displacement of the center of pressure was roughly 11cm. Typically the movement stopped short of the original center of pressure position.

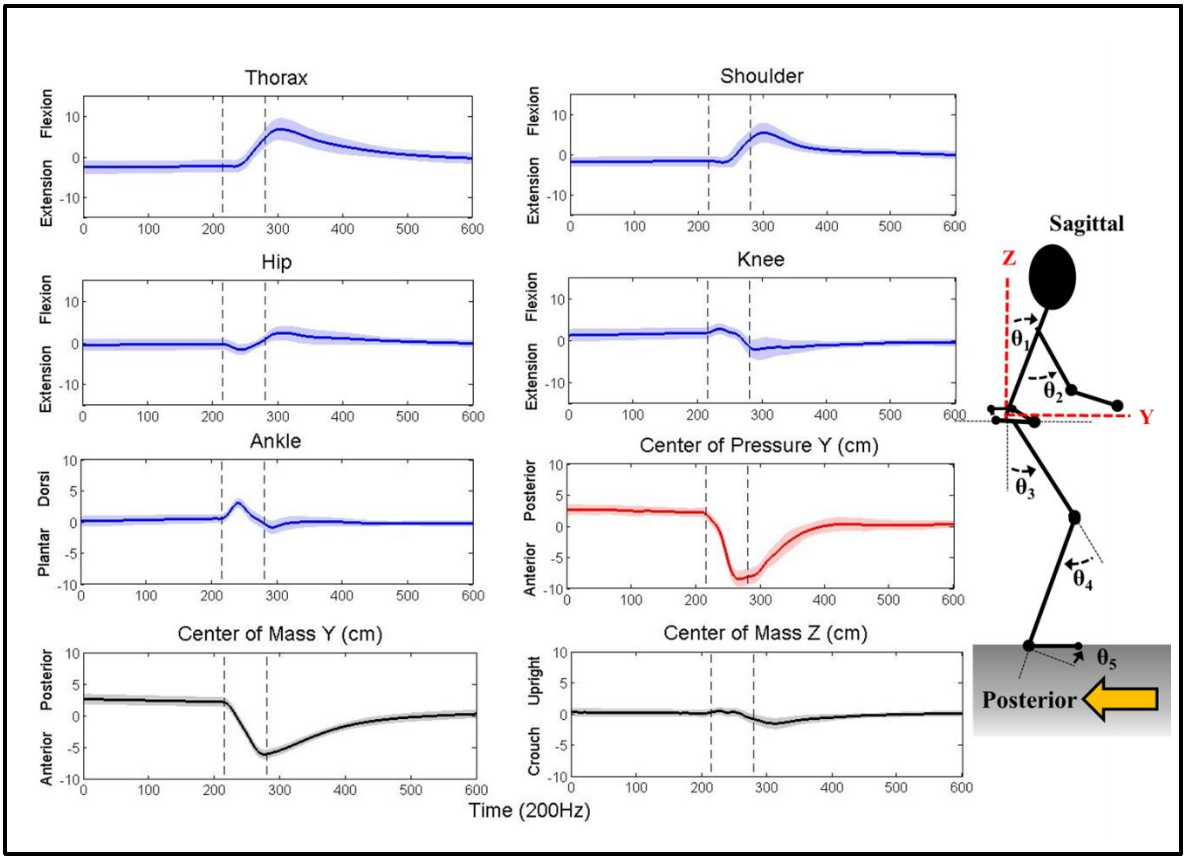

Figure 14: Average kinematic, center or mass, and center of pressure profiles (mean +/- SD) for all subjects during the Posterior perturbation. Vertical lines indicate start and end of perturbation. Note: Initial start position was normalized.

Statistical Results

A series of one-way ANOVA's on each of the nine dependent variables was conducted. The were no kinematic differences, however there was a significant difference in CoPy, $F(1,47) = 6.35$, $p = .015$ and in CoPex, $F(1,47) = 6.76$, $p = .013$ as recreational players displayed greater center of pressure values compared to elite players (Table 7). In terms of the CoPex, a significant interaction was also found. Upon further analysis of this interaction, it was determined that recreational players on skates showed significantly greater CoPex (24.7cm) than both elite players on skates (20.2cm) and recreational players on bare feet (21.5 cm). See Figure 15 for a comparison of the significant differences.

Table 7: Posture displacements (means and standard deviations) of the nine dependant variables during the Posterior perturbation.

Factor		Center of Pressure (cm)		Sagittal Kinematics (θ^o)					Center of Mass (cm)	
		ΔCoPy	CoPex	Thrx	Shldr	Hip	Knee	Ank	ΔCoMy	ΔCoMz
Elite	Mean	**-10.6***	**20.3***	10.5	7.6	5.0	-7.2	-5.1	-8.3	-1.3
	SD	01.1	04.0	04.1	3.7	2.3	2.5	1.7	0.4	0.8
Rec	Mean	**-11.4***	**23.1***	09.0	8.7	5.2	-6.0	-4.8	-8.4	-1.1
	SD	01.2	03.5	03.7	3.4	3.2	2.7	1.7	0.3	1.0
BF	Mean	-10.8	21.0	08.9	7.3	4.7	-6.2	-4.8	-8.3	-1.1
	SD	01.0	03.2	03.9	3.5	2.7	2.3	1.7	0.4	0.9
SK	Mean	-11.2	22.6	10.4	9.2	5.5	-6.9	-5.1	-8.4	-1.3
	SD	01.4	04.5	03.4	3.4	2.9	3.0	1.7	0.4	1.0

*$p < 0.05$, See Figure 14 for indication of (-) direction

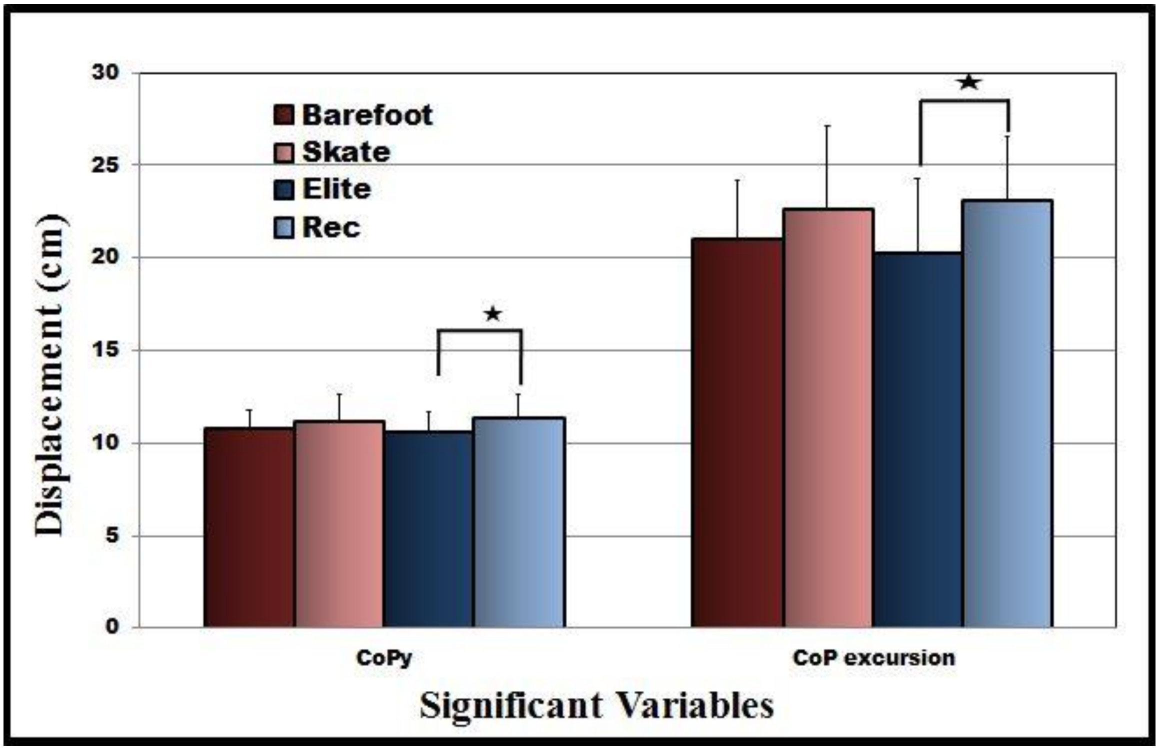

Figure 15: Magnitude of the average center of pressure delta and center of mass excursion values during the Posterior perturbation (+SD bars; p<0.05).

To summarize the responses to the Posterior perturbation, no differences in the kinematic variables were found; however, greater changes in CoPy (11.4 vs 10.6 cm), and in CoPex (23.1 vs. 20.3 cm) between recreational and elite level players was observed. Furthermore, interactions were present where elite players on bare feet showed significantly greater CoPex values (12.9 cm) than both elite players on skates (9.7 cm) and recreational players on bare feet (9.7cm).

3.5 Anterior Perturbation

Perturbation Profiles

The sagittal plane kinematic, center of mass, and center of pressure profiles were similar across calibres and footwear conditions (Figure 16).

Immediately following the Anterior perturbation, the ankle complex plantar flexed by approximately 2º, the knee extended minimally (< 1º), and the hip flexed minimally (< 1º), all passively due to the movement of the force plate. Similar to the Posterior perturbation, during this time the thorax and shoulders were relatively stationary. The CoMy swayed backwards, opposite to the direction of the perturbation, while the CoMz slightly increased as the subject became slightly more upright. Approximately 20 frames (100ms) after the onset of the perturbation, the ankle began to dorsi flex, undergoing roughly a 3º change, while the hip extended by roughly 11º. The thorax, which typically responded after the hip, knee and ankle, tilted 15º posteriorly, while the shoulder flexed by 4º, moving the arms anteriorly. The CoMy and CoMz continued to shift posteriorly (8.5cm), and vertically (4cm), respectively. Similar to the Posterior perturbations, none of the limbs responded by moving back towards their initial position until *after* the perturbation ceased. Approximately 30 frames (150ms) after the platform stopped translating, the opposite movement occurred at each limb in order to bring the body back to its starting position. The ankle slowly plantar flexed, the knee extended, and the hip flexed, all in an attempt to bring the body anteriorly and to move both the CoMy and CoMz back towards their initial positions. During this same time, the thorax flexed, which moved the trunk anteriorly, and the shoulders extended back towards their starting position.

Opposite of what was demonstrated for Posterior perturbations, the Anterior perturbation caused the center of pressure to move posteriorly towards the rear of the stance, corresponding with posterior sway of the CoMy. The center of pressure movement continued until approximately 10 frames (50ms) before the force plate stopped translating, at which time the subject began to recover and shift the center of pressure back towards the front of their stance. The total posterior displacement of CoPy was roughly 10cm. Typically the movement stopped short of the original center of pressure position.

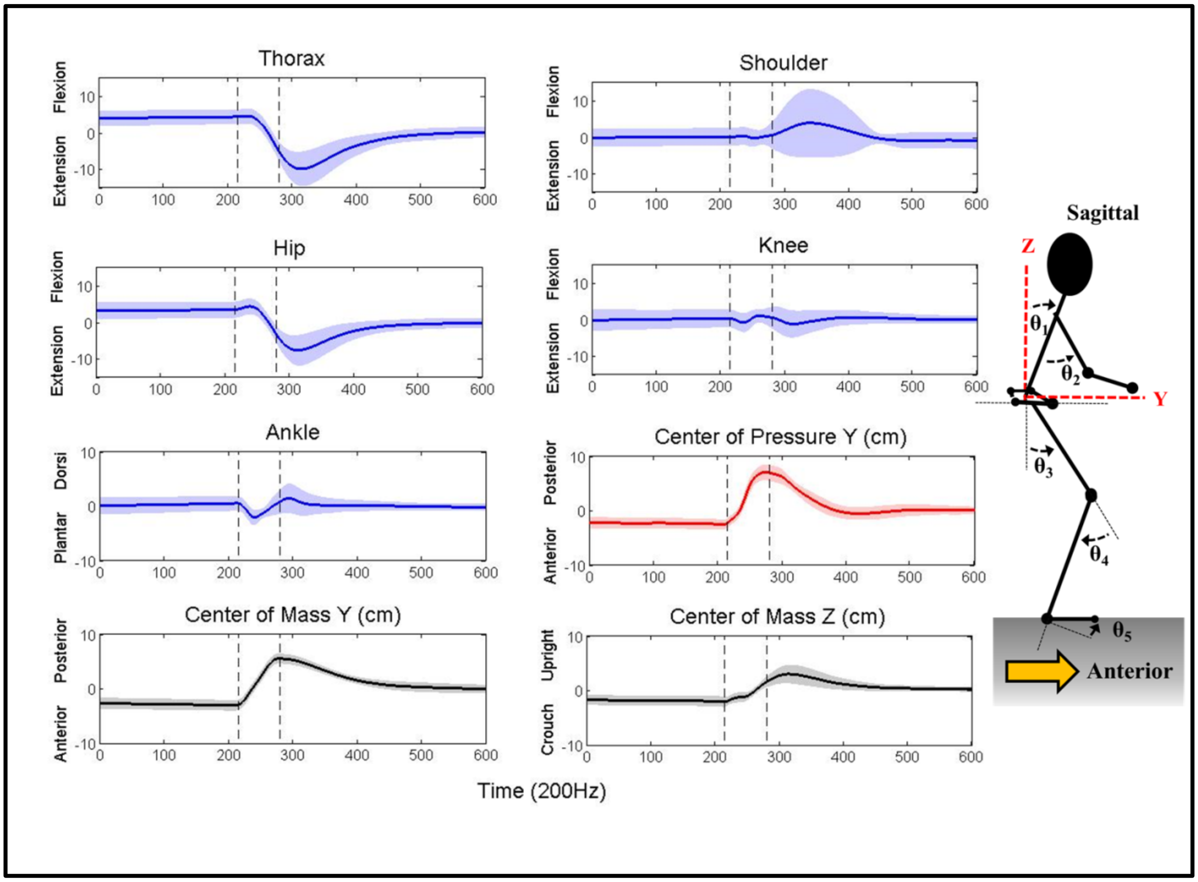

Figure 16: Average kinematic, center or mass, and center of pressure profiles (mean +/- SD) of all subjects during the Anterior perturbation. Vertical lines indicate start and end of the perturbation. Note: Initial start position was normalized.

Statistical Results

A series of one-way ANOVA's on each of the nine dependent variables was conducted. The were no kinematic differences; however, there was a significant difference in CoPy, $F(1,47) = 7.90$, $p = .007$ and in CoPex, $F(1,47) = 11.72$, $p = .001$ between barefoot and skate conditions, as skate conditions yielded greater center of pressure values (Table 8). See Figure 17 for a comparison of the significant differences.

Table 8: Posture displacements (means and standard deviations) of the nine dependent variables during the Anterior perturbation.

Factor		Center of Pressure (cm)		Sagittal Kinematics (θ°)					Center of Mass (cm)	
		Δ CoPy	CoPex	Thrx	Shldr	Hip	Knee	Ank	Δ CoMy	Δ CoMz
Elite	Mean	09.7	18.1	-15.2	8.0	-13.8	-1.6	2.3	8.5	4.8
	SD	01.0	03.2	7.3	14.4	6.6	8.5	5.6	0.5	3.0
Rec	Mean	09.9	20.2	-15.0	1.3	-12.0	-0.1	4.3	8.7	3.8
	SD	01.5	05.3	5.2	13.2	4.9	8.0	5.3	0.5	1.9
BF	Mean	**09.3***	**17.2***	-14.7	3.3	-12.3	0.2	4.3	8.6	3.8
	SD	01.2	01.2	6.6	13.2	6.1	8.4	5.8	0.5	2.2
SK	Mean	**10.3***	**21.3***	-15.5	5.4	-13.4	-1.8	2.7	8.6	4.7
	SD	01.2	04.3	5.8	14.8	5.5	8.0	5.1	0.5	2.7

*Significant difference, See Figure 16 for indication of (-) direction

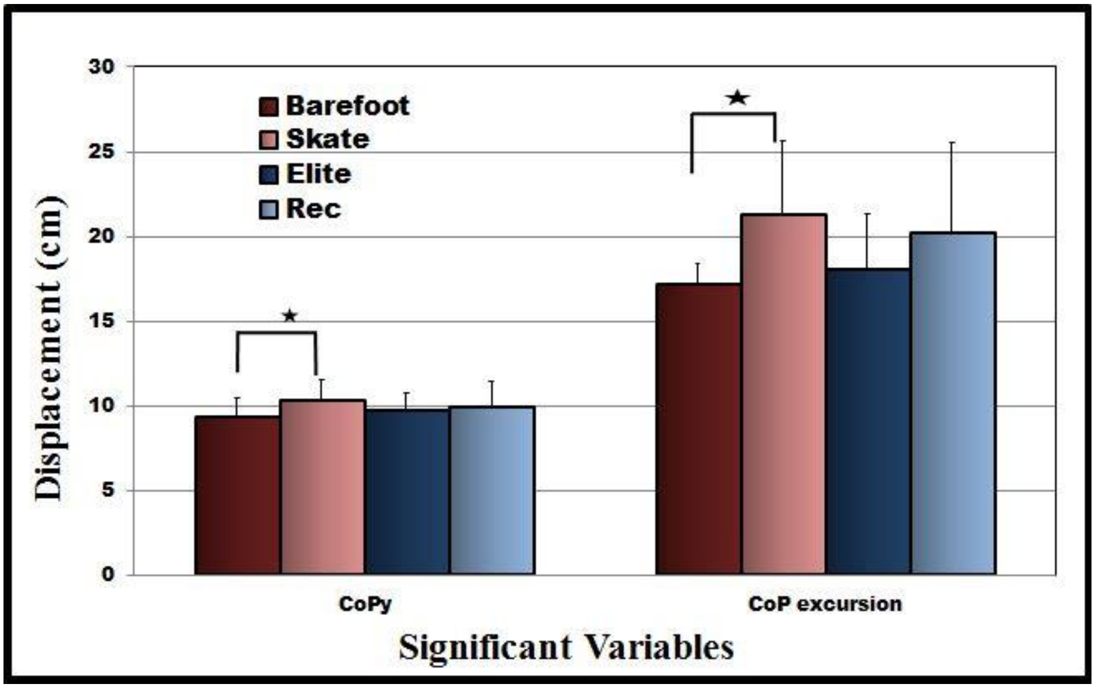

Figure 17: Magnitude of the average center of pressure delta and center of mass excursion values during the Anterior perturbation (+SD bars; $p<0.05$)

Although there were no significant differences in the sagittal plane kinematic and center of mass profiles across calibre and footwear conditions, upon inspection of individual subjects' profiles, some discrepancies were observed. When examining the kinematics of the shoulder, it is obvious that there was a large degree of variability in the response (Figure 16, top right, frames 300 to 400). Upon closer inspection of the individual plots, it was revealed that some subjects displayed large amounts of shoulder flexion (~10º, n = 13) while the remaining subjects demonstrated smaller amounts of shoulder extension (~ -3º, n

= 11), therefore when grouped together, a net response of flexion was displayed. However, the shoulder angle relative to the thorax, the extension may be somewhat misleading. When video footage was examined, it was revealed that although some subjects were undergoing shoulder extension, in reality they were keeping their arms in a stationary position while their thorax extended backwards. Regardless of the response of the shoulder (i.e. flexion or extension), there were no statistical differences across calibres or footwear conditions.

3.5 Functional group classification during Anterior Perturbation

Another result that was revealed after examining individual subjects' profiles was that of the knee response. Specifically, some subjects displayed knee and ankle patterns that were the opposite of what would be expected based on both the postural control literature, and the results of the posterior perturbation. Due to this unexpected and somewhat atypical response, subjects were divided into three distinct functional groups (post hoc) based solely on the kinematic profile of the knee angle and without any regard for player calibre; those that

1) responded primarily with knee flexion (n = 9),
2) responded primarily with knee extension (n = 9), and

3) those that did not display a quantifiable, definitive, or consistent pattern of knee flexion or knee extension (n = 6).

In terms of the three different strategies, the observed differences were found in the knee and ankle kinematics, as well as in the CoMz movement. The first group, demonstrating a knee flexion strategy, had a kinematic response that was expected based on the Posterior perturbation response, and what was found during prior upright studies. The sequence of movements was as follows: immediately following the initiation of the perturbation, the ankle plantar flexed by roughly 2º and the knee extended by approximately 1º, due to the translation of the force plate. Twenty frames (100ms) after the perturbation, the body started to respond to the movement of the force plate, and the ankle dorsi flexed by roughly 5º while the knee flexed by approximately 8º. During this entire time, the CoMz slowly increased (~ 3 cm). These movements continued until about ten frames (50ms) after the force plate stopped translating. At that point, the ankle slowly plantar flexed, the knee extended, and the CoMz decreased, all in an attempt to bring the body back to its original position.

The second group, with unexpected knee extension, displayed conflicting knee and ankle kinematics when compared to the knee flexion group. Immediately following the initiation of the perturbation, the ankle plantar flexed, and the knee started to extend, again due to the translation of the force plate. This was identical to the initial movements displayed by the knee flexion group. **However, the difference became apparent once the body started to respond to the perturbation. Instead of displaying a clear dorsi flexion movement, the ankle either became relatively stationary or only slightly dorsi flexed, by a magnitude of less than 1º. The greatest difference was seen in the knee angle though, as instead of flexing, the knee clearly continued to extend (~7º). This extension also lead to an increase in the CoMz position by roughly 5 cm, an indication that the individual was becoming for upright.**

The third group did not display a clear preference for either knee flexion or knee extension during the perturbation, nor did they demonstrate the consistency that was observed in the previous two groups. In terms of the knee and ankle angles, the third strategy was arguably a hybrid of the first two strategies. See Figure 18 for a profile comparison of these three balance recovery strategies.

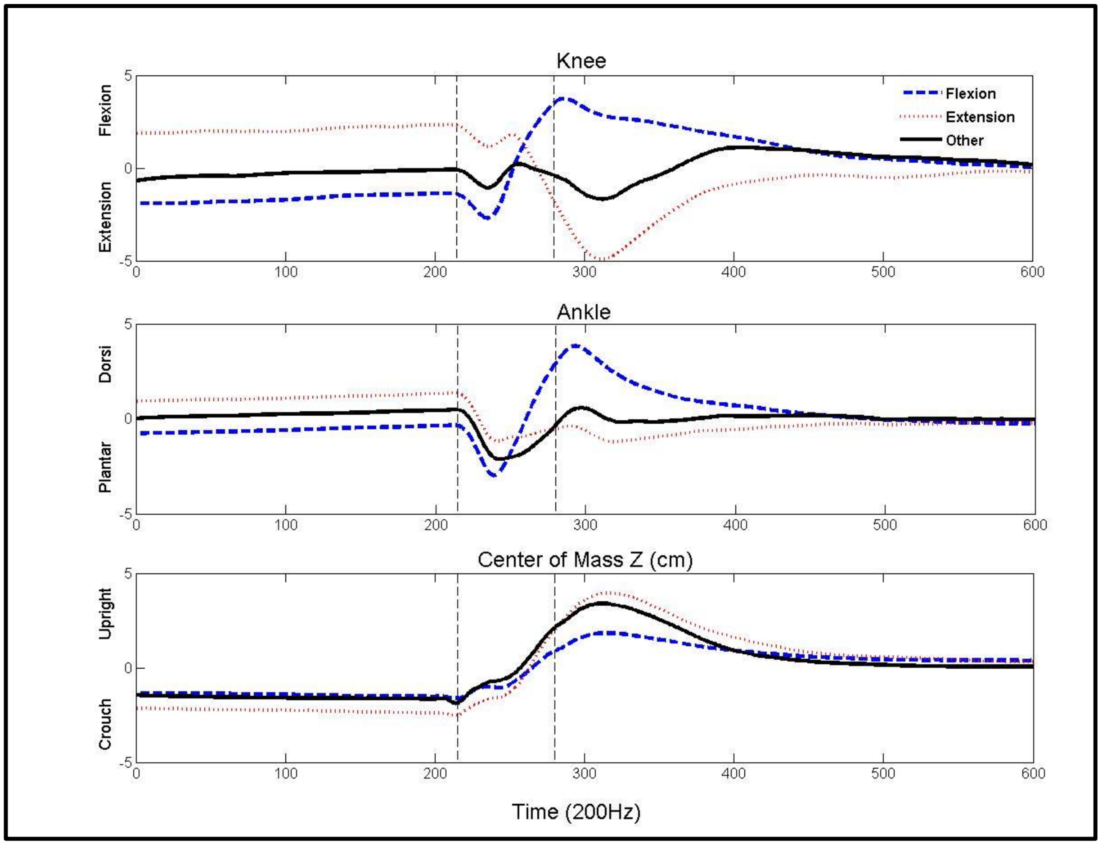

Figure 18: Knee, ankle, and center of mass (z) comparison of the three different knee strategies players demonstrated during the Anterior perturbations.

In terms of the A-P perturbations (Anterior-Posterior), the variables generally showed similar results. By comparing the Posterior perturbation graph with the Anterior perturbation graph you can see that both perturbations showed almost identical trends in CoPy and CoP_{ex}, with a few significant differences observed (Figure 19).

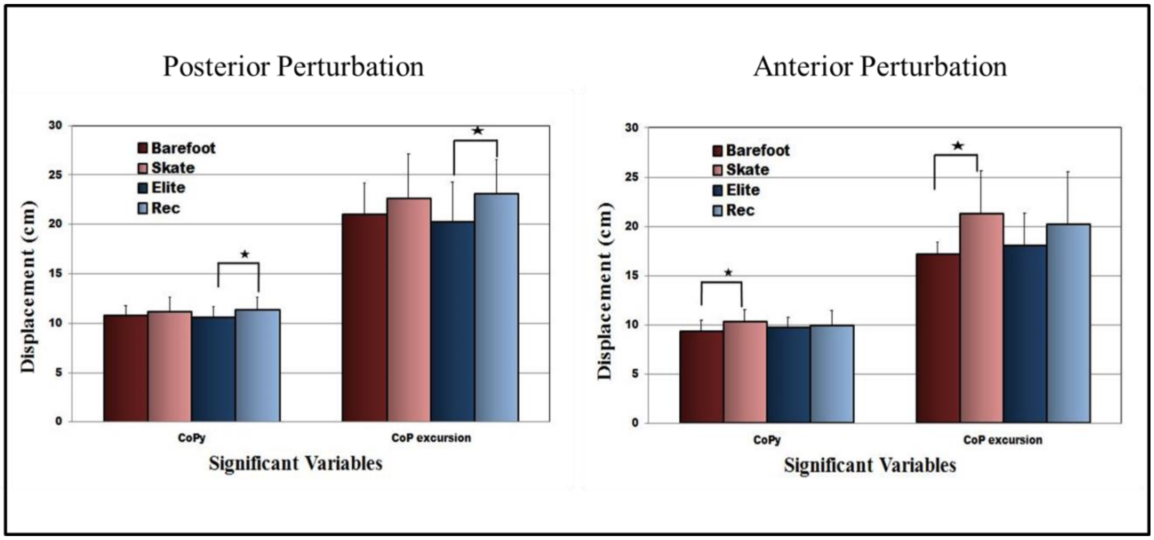

Figure 19: Comparison of trends and significant differences in the Posterior and Anterior perturbations (+SD bars; p<0.05).

To summarize, the Anterior perturbation resulted in differences in the kinetic variables, as skate conditions showed greater changes in both CoPy (9.3 vs. 10.3 cm) and CoPex (21.3 vs. 17.2 cm) compared to barefoot conditions. Although there were no significant differences in the kinematic variables, there were some differences in the shoulder response as well as drastic differences in the knee angle profiles across subjects. Due to these knee angle differences, three post hoc groups were created based on the knee's response to the perturbation, so that additional analyses could be conducted.

3.6 Comparison of Initial Body Position across Balance Strategy Groups

In order to determine other possible covariates related to the knee kinematic response differences, the subjects' initial positions were further examined. Each subject had been instructed to consistently adopt a semi-squat start position prior to each trial; however, subjects were allowed to choose the extent of hip and knee flexion preferred (Figure 4). Feedback was not given to the subjects on their initial joint angles, nor was an exact position required. Consequently, some subjects adopted a position that was deeper into a squat, while other chose a stance that was more upright. Post hoc estimates of initial position were based on hip, knee, and ankle angles, as well as the center of mass height (relative to one's body height) immediately prior to the onset of the perturbation. A series of one-way ANOVA's on each of the latter four dependent variables at initial position were conducted. There was a significant difference across the strategy conditions in terms of initial hip angle, $F(2, 47) = 11.20$, $p = .000$, initial knee angle, $F(2, 47) = 9.22$, $p = .000$, and initial CoM height, $F(2, 47) = 66.30$, $p = .000$. A series of post-hoc analyses (Tukey HSD) were performed on the strategy conditions in order to determine where the individual differences between the three groups occurred. The most relevant results were revealed when the difference between the flexion and extension strategy group was examined, as there was a significant difference in initial hip angles, knee angle, and CoM height. Specifically, the knee extension group had a larger hip and knee angle, and a lower center of mass position compared to the knee flexion group. See Table 9 for the means and standard deviations of each variable.

Table 9: Mean and standard deviation of the four dependent variables used to assess a subjects' initial body position

		Sagittal Kinematics (θº)			Center of Mass (%)
Factor		Hip	Knee	Ankle	CoMz
Flexion	Mean	**46.3**[2]	**37.5**[2,3]	15.5	**54.1**[2]
	SD	06.5	09.2	05.6	02.7
Extension	Mean	**60.1**[1,3]	**48.1**[1]	17.5	**51.7**[1,3]
	SD	10.0	05.7	05.3	03.0
Other	Mean	**50.9**[2]	**45.2**[1]	17.4	**54.5**[2]
	SD	10.7	08.7	02.6	02.3

[1]$p < 0.05$ between flexion group

[2]$p < 0.05$ between extension group

[3]$p < 0.05$ between other group

[*] $p < 0.05$

Ultimately, this demonstrates that the group of subjects that responded to the perturbation primarily by extending their knee were significantly lower in the squat position, in terms of hip (14º), knee (10º) and CoMz (2%), compared to the group that responded primarily with knee flexion, as depicted in Figure 20.

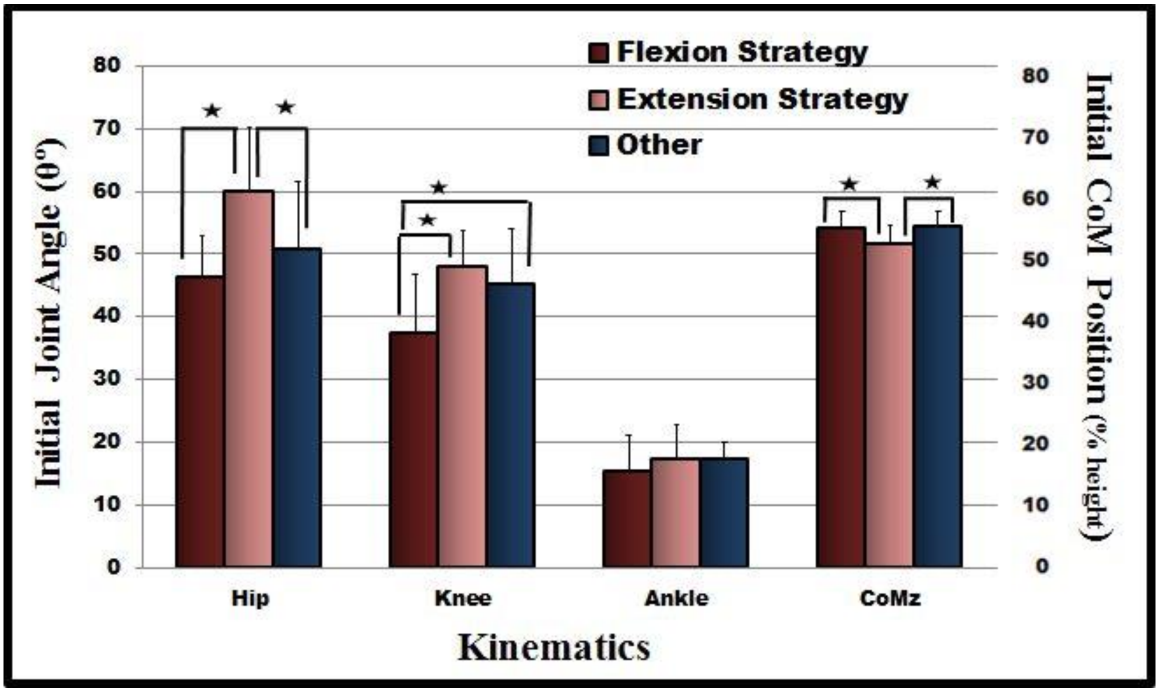

Figure 20: Average initial angle of the hip, knee, ankle, and the center of mass (z) between the three balance strategies (+SD bars; $p<0.05$).

3.6 Statistical analysis of the perturbation response

In order to determine whether these different strategies were in fact statistically different in terms of how they were functionally group (knee flexion/extension), a final analysis was conducted using the same dependent variables that were previously tested; however, the calibre condition was replaced with the newly created strategy condition. Similar to the original analysis performed on each of the four perturbation conditions, a series of one-way ANOVA's on each of the nine dependent variables was conducted. Significant differences were observed across the strategy conditions in terms of the change in knee angle, $F(2, 47) = 56.72$, $p = .000$, change in ankle angle, $F(2, 47) = 14.74$, $p = .000$, and change in CoMz position, $F(2, 47) = 10.70$ $p = .000$. However, no significant difference in hip angle was found. A series of post-hoc analyses (Tukey HSD) were performed on the strategy conditions in order to determine where the individual differences amongst the three groups occurred. Again, the most relevant results revealed significantly different changes in knee angle, ankle angle, and CoM height between the flexion strategy and the extension strategy (Table 10).

Table 10: Means and standard deviations of the nine dependant variables grouped according to their kinematic balance strategy (flexion, extension, or other).

		Center of Pressure (cm)		**Sagittal Kinematics (θ°)**					**Center of Mass (cm)**	
		CoPy	**CoPex**	**Thrx**	**Shldr**	**Hip**	**Knee**	**Ank**	**CoMy**	**CoMz**
Flex	Mean	09.8	19.7	15.5	-00.1	11.1	**7.1**[2,3]	**7.6**[2,3]	8.5	**2.7**[2]
	SD	01.1	04.4	06.3	06.3	06.0	4.0	3.8	0.4	1.8
Ext	Mean	09.9	19.3	13.6	04.2	14.4	**-8.4**[1,3]	**-0.3**[1]	8.6	**5.8**[1]
	SD	01.5	05.2	07.1	17.8	06.1	4.5	4.8	0.5	2.4
Other	Mean	09.5	18.4	16.7	11.2	13.0	**-1.2**[1,2]	**3.2**[1]	8.7	4.2
	SD	01.2	04.1	04.2	14.2	04.5	5.6	4.3	0.5	2.0

[1]p< 0.05 between flexion group

[2]p< 0.05 between extension group

[3]p< 0.05 between other group

Ultimately, these results show that one group adopted a knee flexion approach while the other group adopted a significantly different knee extension approach. However, there were no significant difference across strategy groups in terms of the kinetic variables, and therefore the center of pressure response was similar regardless of the strategy used. See Figure 21 for a comparison of the significant differences.

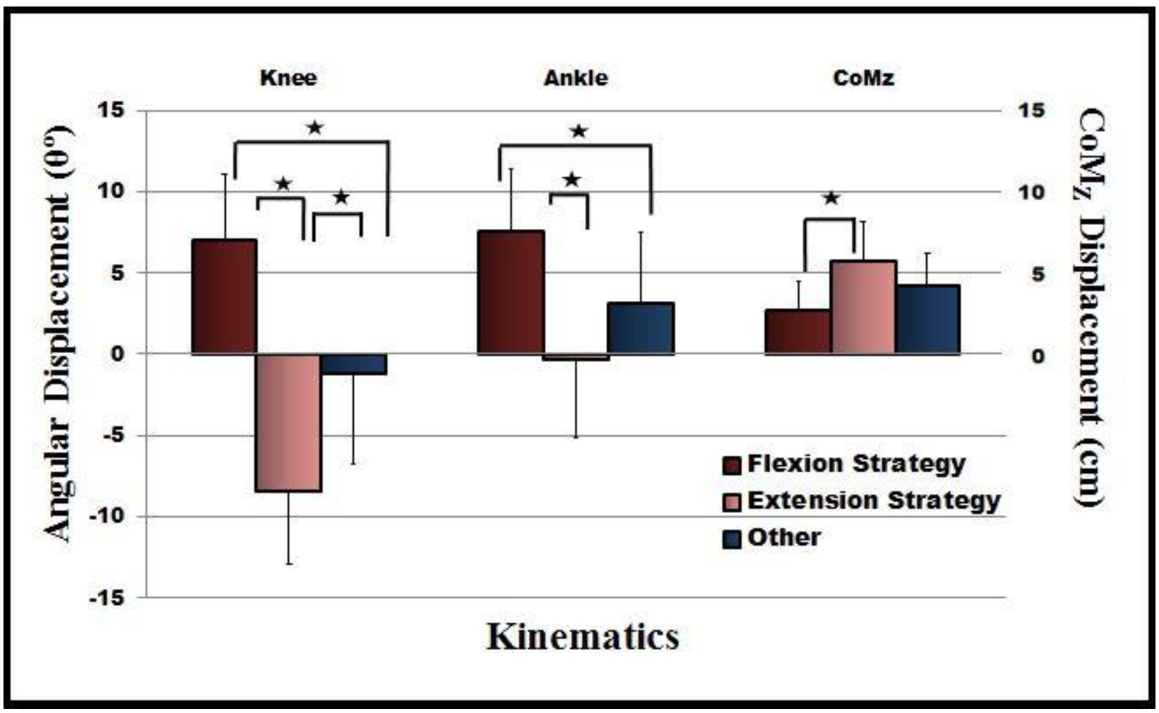

Figure 21: Comparison of both the average angular displacements of the knee (+ flexion, - extension) and ankle (+ dorsi, - plantar) and the vertical displacements of the center of mass between the three balance strategies (+SD bars; p<0.05).

Chapter 4: Discussion

The goal of this study was to determine how wearing ice hockey skates would affect static stability (compared to without) as well as whether one's ice hockey playing calibre modifies standing stability. To examine this, quantitative measures of body kinematics, center of pressure, and center of mass responses were measured during surface perturbations in both footwear conditions. Conventional wisdom would suggest that wearing skates would compromise stability righting responses to perturbations, and that higher player caliber would convey more adroit responses than recreational players. Differences were found between footwear and calibre conditions; however, not necessarily as predicted. The following text provides an interpretation of the quantitative measures obtained.

4.1 Kinematics

Footwear and Calibre

In general, the kinematics responses across both footwear and calibre conditions were remarkably similar. Group kinematic profiles within each perturbation showed comparable dynamic response patterns and displacement magnitudes (with the exception of the Anterior perturbation: see 4.4). This was not expected based on pilot testing results wherein drastic instability was observed in skates versus the more stable barefoot condition. Indeed, so unstable was the skate condition for the lower calibre players that balance was difficult to maintain without stepping; hence, for the final testing protocol the initial start position of **upright stance** was replaced with a “**hockey ready**” **stance** (or alternatively described as a “**crouched**” posture) so that all subjects would be able to complete all perturbation tests, as well as to offer a more externally valid approach. Unintentionally, this change in the final protocol’s start position posture proved inherently much more resistant to Anterior and Posterior (A-P) perturbations than expected; that is, in the crouched posture the lower limbs’ joints were optimal positioned to act as a passive dampening spring.

With regards to the Ipsi and Contra (M-L) perturbations, the protocol set the subjects’ stance width normalized to 1.15 times the distance of the shoulders (i.e. between the left and right acromioclavicular joints); again to mimic typical on-ice stance. Evidently this stance width afforded a sufficiently stable medial-lateral base of support that again minimal differences were found between skate/barefoot and elite/recreational calibre players. In comparison to the A-P stance width (i.e. base of support ~ 30 cm), the average mediolateral stance width of 40cm was 33% greater, therefore offering a more stable M-L base of support.

4.2 Center of Mass

Footwear and Calibre

For the Ipsi and Contra perturbations, subjects on skates exhibited greater CoMx sway (~1 cm) than in the bare foot condition, while conversely in the static condition subjects on skates displayed greater variability in their CoMx sway. In the Anterior and Posterior perturbations, no differences in CoMx sway were found.

Although these differences were statistically significant, its small magnitude (~ 1cm) may be of little practical value when standing on two feet. However, in single leg stance (as transiently experienced in skating) a 1 cm greater M-L sway due to a perturbation would have a profound effect as the base of support would be reduced from ~40 cm to less than 0.3 cm (i.e. the width of the blade). Whether these bilateral stance findings are telling of single leg balance remains to be determined in future studies. If this greater CoMx sway is meaningful, it may be explained in part by the differences in the ankle complex's height above the ground surface. Due to the narrow skate and blade holder that extends from the boot, the ankle/foot complex in the skate becomes inherently more unstable about the medial lateral axis, when compared to the bare foot in that same axis (Figure 22).

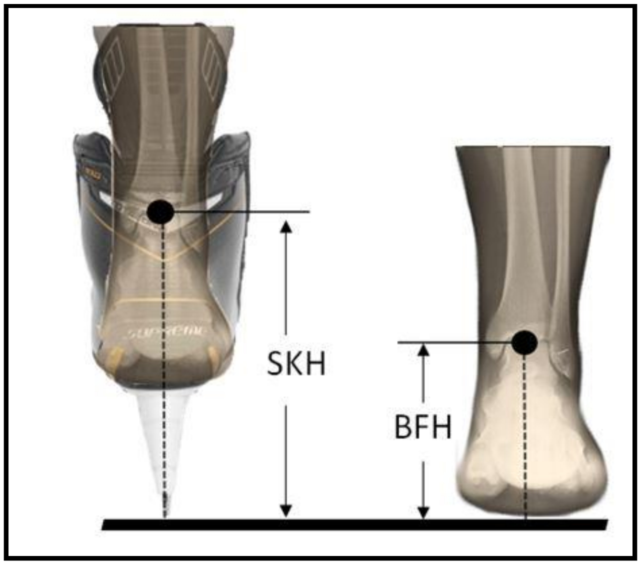

Figure 22: Comparison of the height differences of the ankle complex between skate and barefoot conditions (SKH 6.3 cm > BFH).

The same CoMy sway differences **were not** seen in the A-P perturbations, possibly because skate blade and foot length are more similar.

The current CoM sway values were comparable to those found in previous studies using platform perturbations. Henry, Fung, and Horak (1998) observed CoM changes of 6 cm for posterior perturbations (vs 8.3 cm), and 5 cm for Anterior perturbations (vs 8.6 cm). The increase in barefoot CoMy sway observed in the current study may be attributed to using a slightly greater perturbation magnitude than the perturbation used by Henry et al. (1998) (i.e. 10cm in 250ms vs 9cm in 300ms).

4.3 Center of Pressure

Footwear

Significant differences in the body's center of pressure's maximum displacement or total excursion distance emerged in all four perturbation conditions. During Ipsi and Contra perturbations, $CoP_{BF}>CoP_{SK}$, whereas during Anterior and Posterior perturbations, $CoP_{BF}<CoP_{SK}$. These results appear to be counter intuitive; however, it may be explained by the different relative foot (or skate) / floor contact dimensions. The CoP on bare feet was measured at the foot-plate interface where the entire width of the feet were in contact with the force plate. On the other hand, the CoP on skates was measured at the blade / force plate interface where only the width of the skate blade was in contact with the force plate. During M-L perturbations the majority of the movement occurs in the frontal plane and therefore CoPx was most affected. Along the X axis, the width of the foot is much greater than the width of each skate blade (10.3 cm vs. 0.3 cm) and therefore each foot affords the potential for more CoPx movement without sacrificing ground contact stability (Yu & Tu, 2009). During the A-P perturbations CoPy was most affected. Along the y axis, the length of the foot and the blade delimit the range of CoPy movement. The blade being profiled (i.e. curved) and longer than the sole of the foot (30 vs. 23.7 cm) affords a

greater “rocker” in comparison to standing on bare feet (Yu & Tu, 2009) as depicted in Figure 23.

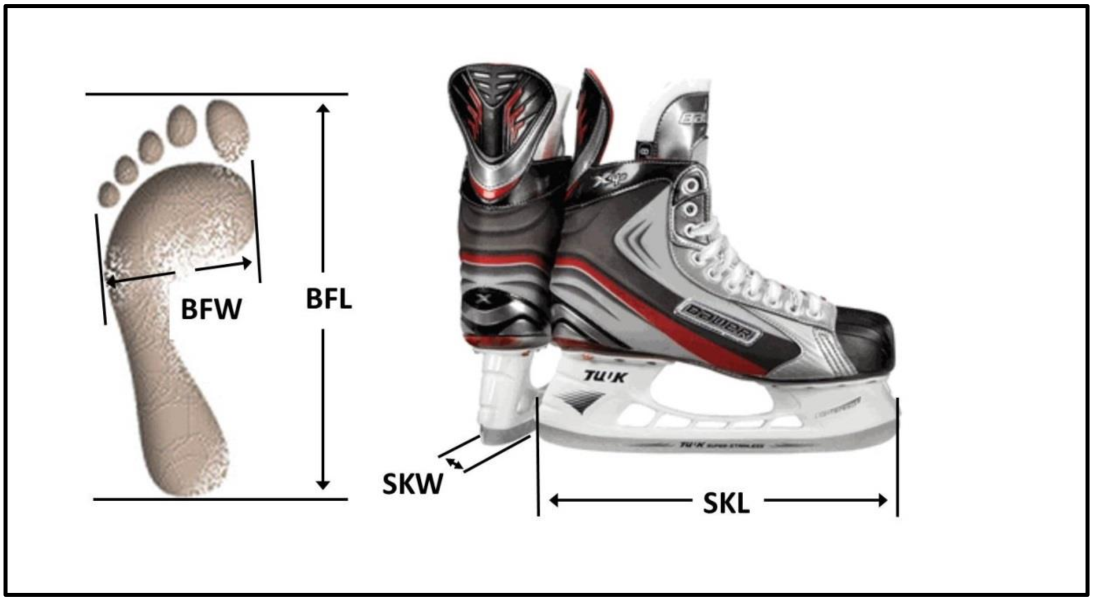

Figure 23: Comparison of the skate blade length (SKL = 30cm), skate blade width (SKW = 0.3 cm), barefoot length (BFL = 23.7cm), and barefoot width (BFW = 10.3 cm). (Yu & Tu, 2009)

Barefoot CoP values were similar to those obtained in past studies. Henry et al. (1998) observed CoP delta values for posterior perturbations of 9 cm (vs. 10.8 cm) and anterior perturbations of 7 cm (vs. 9.3 cm). Similar to the CoM changes, the increase we have observed may be attributed to the increase in the perturbation magnitude.

Calibre

Significant differences in the center of pressure variables were found between elite and recreational players during all four perturbations. During the Ipsi and Contra perturbations, elite subjects showed greater CoPx delta values, whereas in the Anterior and Posterior perturbations, recreational subjects showed greater CoPy delta values, and greater CoP excursion. In terms of the former result, most studies generally have found the opposite, i.e. $CoPy_{ELITE} < CoPy_{REC}$ (Era et al., 1996; Konttinen et al., 1999; Niinimaa & McAvoy, 1983). However, these differences may be explained by the level of difficulty of the tasks (i.e. the ratio of CoP to respective directional base of support). This ratio is much smaller in the M-L direction than the A-P direction. Hence, the challenge to stability is less in the M-L direction, thus skill level was not as crucial of a factor.

Supporting that, qualitative observations as well as subjects' verbal reports, indicated that it was more difficult to maintain balance during the A-P perturbations compared to the M-L perturbations. These observations were also reflected in the quantitative data for the A-P perturbations. For instance, the overall movement of the CoM and the Thorax (largest body segment recorded) were greater during A-P than during M-L perturbations, in terms of both CoM sway (8.5 vs. 7.4 cm) and Thorax displacement (12.5 vs. 4.2 cm). Therefore, if the assumption that the M-L perturbations were easier to manage, players may not have been challenged enough to elicit a detectable difference between skill level. Similar to previous studies, it has been shown that during barefoot static conditions and barefoot M-L instability conditions (i.e. less challenging) elite national level skiers demonstrate “worse” postural control than regional skiers (i.e. greater CoP surface area and greater CoP velocity), while the same result was not observed in the A-P condition (Noé & Paillard, 2005). However, the authors attributed this difference to the fact that elite skiers were familiar to the rigid support that a ski boot provides, and not the fact that the task was less challenging. However, in a study conducted on elite skiers using only A-P directed perturbations (i.e. more challenging) the elite group displayed smaller and less variable CoM velocity than the control group (Asaka et al., 2012). In a study on the postural control of surfers, it was shown that in static postures (i.e. less challenging), national surfers did not perform differently than regional surfers; however, in more challenging postures, such as during M-L and A-P instabilities, the national surfers displayed greater postural control (i.e. smaller CoP excursion and velocity) (Paillard et al., 2011). Furthermore, in that particular study, subjects stood with their feet together, therefore it can be argued that the M-L level of difficulty was just as great as the A-P difficulty since the base of support had comparable dimensions both in the X and Y direction. Ideally, one would test multiple levels of difficulty (i.e. increasing perturbation magnitudes and/or decreasing base of support) in a more incremental way, in order to determine if the level of postural control difficulty has an effect on the performance of subjects' of different skill levels.

4.4 Flexion – Extension strategy

As noted in the results, on close inspection of the data, distinct differences in the knee response during the Anterior perturbations were observed. Some subjects flexed the knee, others extended, and yet others displayed a hybrid response. Why this break from one general trend seen in the other perturbation directions? What was causing these polar opposite knee responses to the same Anterior perturbation? We believe this interesting finding stems from the use of a crouched initial posture.

In the current study, analysis of the initial body positions revealed that the knee extension group was significantly deeper into the squat position (14º greater hip flexion, 11º greater knee flexion, and 2% lower CoMz) immediately prior to the perturbation (Figure 24b) compared to the typical knee flexion strategy group (Figure 24a). Therefore, it was suspected that the player's initial position adopted prior to a perturbation could determine in some capacity the kinematic knee response that allows subjects to maintain balanced. Interestingly, in the extensive body of literature on standing postural control, most studies have had subjects starting in the anatomical position (figure 24c) and examined factors such as stance width, single leg support, support surface material, footwear design, the effect of vision, or the effect of age (Alexander, Shepard, Gu, & Schultz, 1992; Alpini et al., 2008; Henry et al., 2001; Lamoth & van Heuvelen, 2011; Noé & Paillard, 2005; Paillard & Noé, 2006; Riemann, Myers, & Lephart, 2003) but few studies have considered the effect of a crouched stance. Only Noé and colleagues included a knee flexed start posture in their study of the effect of a restrictive alpine ski boot on postural control. Their protocol had subjects starting with a large amount of knee flexion (Noé et al., 2009). Using a goniometer, subjects adopted a vertical trunk position with 100º (+/- 5º) of knee flexion, in order to mimic one of the postures typically observed during skiing (Berg & Eiken, 1999). However, this study did not look at the dynamic response following a perturbation, nor did it measure whole body kinematics.

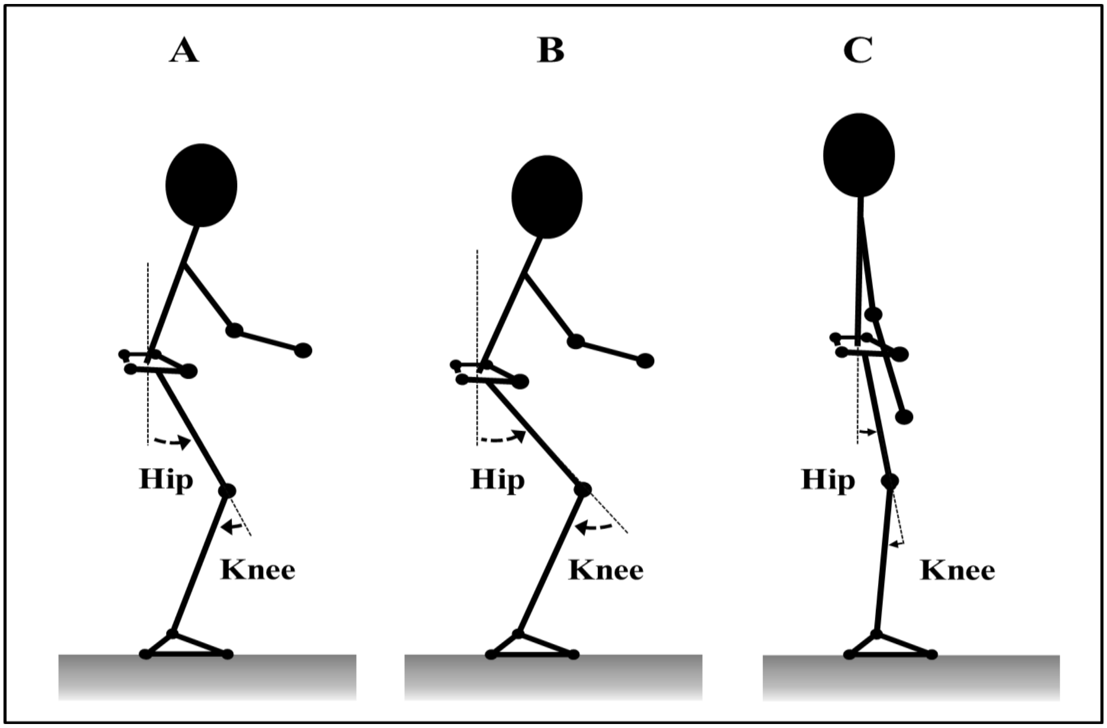

Figure 24: Representation (not to scale) of the initial hip and knee position adopted by the (A) knee flexion group; (B) knee extension group; (C) previous postural control studies.

One of many reasons why most of these previous studies used an erect posture, is that it is the most prevalent form of standing. Crouched stance is more commonly associated with pathologies (i.e. cerebral palsy muscle tendon contractures). For example, children with cerebral palsy often walk with a crouched gait, defined by an exaggerated hip and knee flexion (Damiano, Arnold, Steele, & Delp, 2010; Delp, Arnold, Speers, & Moore, 1996). In our case, the benefit of using a posture similar to most of the literature had to be compared to the loss of external validity, as hockey players virtually never adopt a completely upright posture. Needless to say, a better option would have been to conduct the same study using both a typical upright standard position, as well as a preferred hockey stance position. Future studies should address this.

Nevertheless, the fact that the effect of initial body position on postural control and balance has never been studied is quite surprising, considering the potential coaching ramifications it could have. In many sports today, athletes are instructed to “get low”, or “bend their knees”, to become more stable, however no one has addressed the question as to how low. When attempting to box out an opponent in basketball, in what hip and knee position are you most stable? If you were about to receive a hit on the offensive line in football, how low should your center of mass be in order to optimize your chances of staying on your feet? Obviously, many other factors are going to contribute to the outcome and answer of both of these scenarios; however, the athletes’ initial position is going to play an important role. Furthermore, the crouched posture that is commonly seen in many sports today (i.e., alpine skiing, wrestling, American football) may have as much to do with offering a powerful propulsion potential due to lower limb extension as to maintaining balance following some form of perturbation. Regardless of the reason for assuming this initial position, the implications are profound, and highlight the need for future study.

In the present study, the extension group did in fact have a significantly lower initial CoMz position; however, it did not translate into any balance differences, as the CoMy sway, thorax movement, and CoPy changes were similar to those in the flexion group. The initial position did however change the dynamic response of the knee, ankle, and CoMz during the perturbation.

The change from a primarily knee flexion response to a primarily knee extension response leads us to believe that there is a critical threshold initial position of the hip and knee at which the change in strategy occurs. We could hypothesize that this threshold lies directly in the middle of the initial positions of the two observed strategies, which would result in a critical hip angle of 53.2º and a critical knee angle of 42.8º. In terms of the “other” strategy group, the average initial hip (50.9º) and knee (45.2º) angles observed were fairly close to these threshold values. Taking into account that: (a) these angles are similar, and (b) this “other” strategy group did not display a consistent and quantifiable tendency for knee flexion or knee extension, there is support for this critical threshold hypothesis, as depicted in Figure 25.

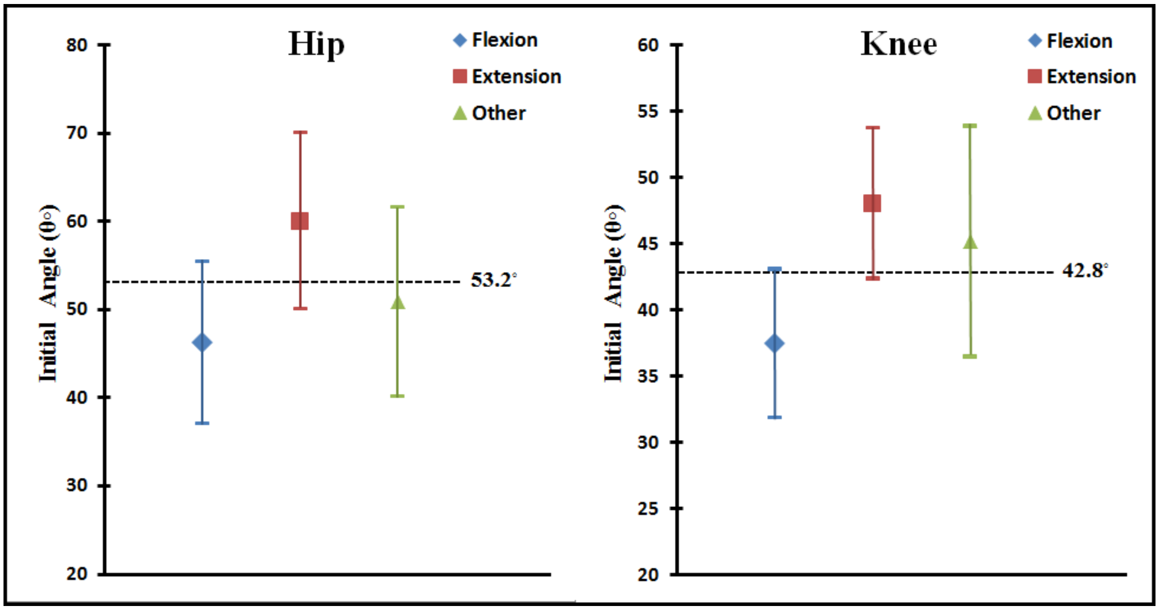

Figure 25: Comparison of the initial hip and knee angle (+/- SD) across the three knee response strategies. Horizontal dashed line represents respective critical values.

However, this critical angle theory is speculative. Ideally, a spectrum of controlled and standardized hip and knee angles would be tested in order to determine more specifically the threshold at which the response of knee flexion transitions to knee extension.

Although limited, there are studies that have found a similar knee extension phenomenon. Stewart, Postans, Schwartz, Rozumalski, and Roberts (2008) investigated the action of the hamstring muscles while in the crouch position using functional electrical stimulation. Subjects had their muscles stimulated while standing in various degrees of crouch, and the kinematic effect was measured. Results indicated that in an upright posture, the hamstrings acted as expected, and flexed the knee upon stimulation. However, as the degree of crouch increased (greater knee angle) the amount of knee flexion decreased and even reversed, producing a knee extension response following stimulation. At 40º of knee flexion, four out of the five subjects demonstrated knee extension, while at 60º of flexion, all five subjects demonstrated knee extension. Because the knee angles between 40 and 60 degrees were not tested, the exact angle at which the transition to extension occurred, cannot accurately be determined. However, the hypothesis of a threshold knee angle of 42.8º is within that range of potential values (Figure 26).

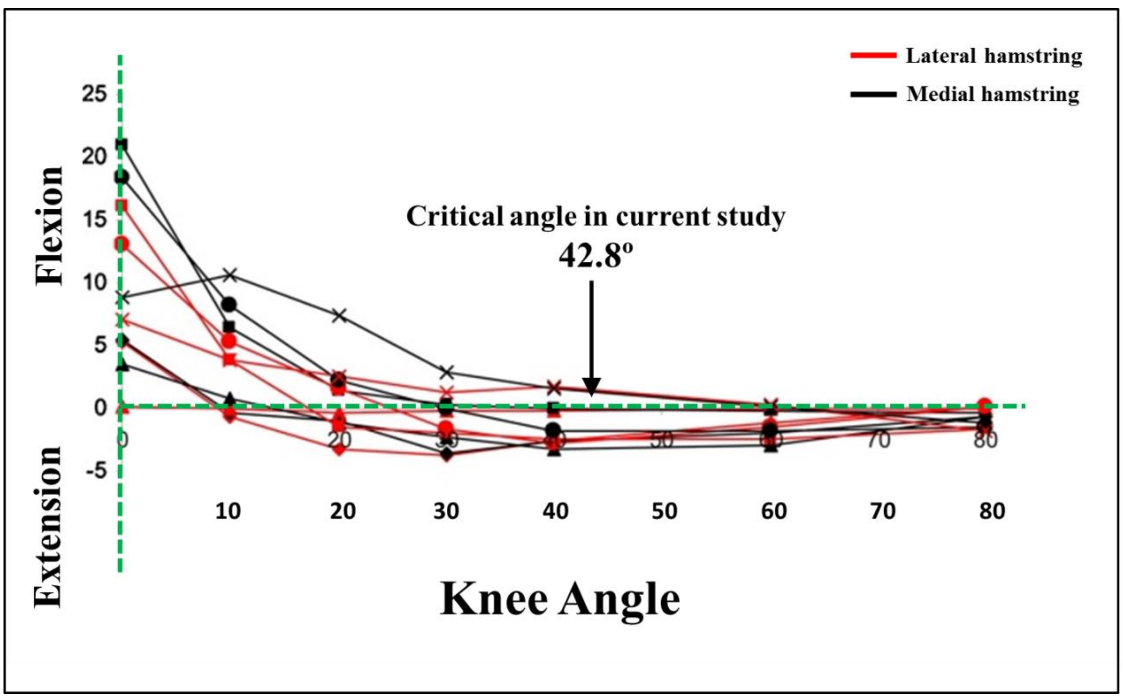

Figure 26: Magnitude of the knee response (flexion vs. extension) due to functional electrical stimulation of the hamstrings during varying degrees of crouched stance. Lines represent individual subjects (Adapted from Stewart et al., 2008)

The study conducted by Stewart et al. (2008) showed that the hamstrings, although generally thought of as a knee flexor, play an important role in extending the knee in crouched postures. Similarly, in a study conducted by Hicks, Schwartz, Arnold, and Delp (2008), a three dimensional model was used to determine the capacity (º/s/N) of muscles used to extend the hip and knee during crouched gait. They found that even in crouched postures, the hamstring muscles were able to maintain their knee extension capacities, whereas all other central hip and knee extensors were not.

Based on these studies, it appears that the hamstrings play a vital role in the postural control transition from knee flexion to knee extension that was observed in the current study. However, the exact cause remains unknown. Evidence suggests that an important factor is that the hamstrings (semitendinosus, semimembranosus, and long head of the biceps femoris) are a group of biarticular muscles, meaning they span both the hip and the knee joint

(Levangie & Norkin, 2001, p. 346). When the orientation of the thigh and shank changes (i.e. during hip and knee flexion) the moment arms at each joint also change, therefore the mechanical advantage of the muscle is modified. Ultimately, when subjects are in a crouched vs. erect posture the mechanical advantage that the hamstrings have on the net knee moment may change. For example, the gastrocnemius, also a biarticular muscle, produces an (a) knee flexor torque and ankle extensor torque. However, it is also able to simultaneously accelerate the (b) knee into extension and ankle into extension, or (c) the knee into flexion and the ankle into flexion (Zajac, 1993; Zajac & Gordon, 1989). All three kinematic combinations are theoretically possible; however, the gastrocnemius cannot accelerate the (d) knee into extension and ankle into flexion (Figure 27). This fourth situation is impossible because as a biarticulate muscle, the gastrocnemius “cannot accelerate both the joints it spans (knee and ankle) simultaneously opposite to its two joint torques” (Zajac, 1993).

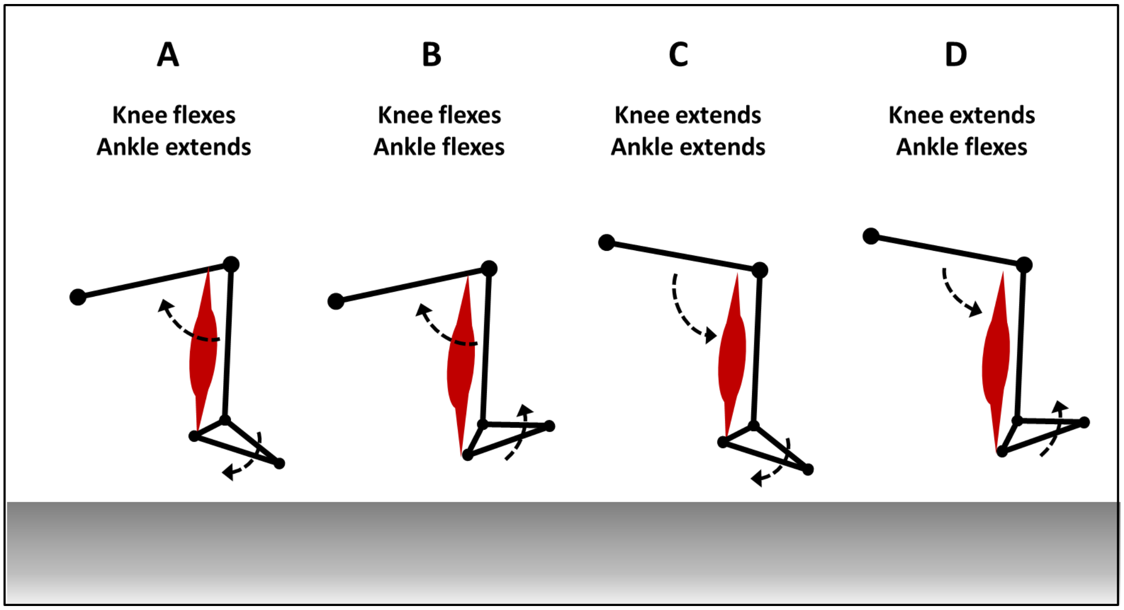

Figure 27: Possible angular accelerations of the knee and ankle joint (A, B, C) produced by the biarticular gastrocnemius. The fourth situation (D) is not possible (Adapted from Zajac, 1993).

The example of the gastrocnemius sheds light into the discussion on the function of the hamstrings muscles, and how they can produce knee extensor moments. Similarly, the biarticular hamstring muscles produce a hip extensor torque and a knee flexor torque, which is what was observed in the subjects that demonstrated the knee flexion strategy. However, two other joint accelerations are also possible based on Zajac's (1993) principles. The hamstrings can also (a) accelerate the hip into flexion and the knee into extension, or (b) accelerate the hip into extension the knee into extension. The latter scenario is what we observed in the subjects that demonstrated the knee extension strategy. Futhermore, in a study of muscle coordination in the squat jump, uniarticular muscles were shown to be the prime movers and provide most of the power, whereas biarticular muscles (i.e. hamstrings) fine-tune the coordination (Bobbert & van Ingen Schenau, 1988). If this fine-tuning characteristic carries over to actions other than the squat jump, such as crouched postural control during an external floor perturbation, it would offer more support for our hypothesis that the hamstring muscles are the important muscle group in the current study.

However, what actually causes the hamstrings to transition in function from flexing the knee to extending the knee when in a crouched posture remains to be seen. Furthermore, the reason why a similar contrasting knee response was not seen during the Posterior perturbation is unknown. Future studies should repeat the current procedures, however strictly monitor and adjust subjects' initial position.

In terms of finding an optimal hip and knee angle, future research would have to be conducted. However, it should be noted that subjects in both the knee flexion and knee extension groups demonstrated similar postural control abilities (i.e. CoP and CoM changes) and therefore a "superior" strategy did not emerge. Because one strategy resulted in a more upright posture (knee extension group), while the other resulted in a slightly more crouched posture (knee flexion group) it would be interesting to introduce a second perturbation immediately following the first one and observe how subjects' respond. Would their "new" initial position would determine how they respond the second perturbation? Would it magnify their kinematic and kinetic response as they would now be less stable than they were prior to the first perturbation? These questions remain.

The introduction of a second perturbation would have implications for sports in which multiple perturbations occur back-to-back. In these cases, an athlete may be able to withstand the initial perturbation (as we observed in the current study); however, based on their new body position they may now be in a better (or worse) position in terms of managing the following perturbation. For example, a sport like downhill skiing, which adopts a crouched posture, may be affected by this phenomenon. If a downhill skier hits consecutive moguls (i.e. perturbations), their initial response is going to be affected by their hip and knee angle, which in turn will affect how low to they respond to the second mogul. Ultimately, the degree of crouch in one's initial posture has the ability to dictate the outcome of an expected surface perturbation within a laboratory setting, and it is believed that this would translate to a sporting environment.

Chapter 5: Conclusion

In summary, the lack of kinematic differences across both footwear test conditions and subject calibres was surprising; likely the adoption of a crouched initial stance proved more crucial to diminish the perturbation disturbances. Significant center of pressure and center of mass changes emerged between both (a) calibre and (b) footwear conditions, however they differed according to the type of perturbation (i.e. M-L or A-P). The latter was possibly due to the different degrees of challenge that each perturbation direction presented, while the former may be attributed to the different skate blade/foot dimensions. The most intriguing result from the entire study was the three distinct knee responses that emerged as a result of the Anterior perturbation. The initial degree of crouch appears to determine the response of the knee, as at some critical hip-knee angle, the dynamic response changed from knee flexion to knee extension. The hamstrings appear to be the muscle group primarily responsible for this transition; however, the exact threshold angle remains unknown, as does the biomechanical or neuromuscular reason for this transition.

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Appendix I – Pre Screening Questionnaire

Name: ___________________________

Age: ___________________________

Hockey Experience (years): ___________________________

Highest Level Played: ___________________________

Current Team: ___________________________

1. In the past year have you suffered any knee or ankle injuries? Has it prevented you from playing hockey? Please explain.
2. In the past year have you experienced any other lower body injuries? (E.g. broken bones, torn ligaments, etc.) Have they prevented you from playing hockey? Please explain.
3. In the past year have you suffered any nervous system injury? (E.g. Damage to a nerve, numbness or pins and needles, etc.) Has it prevented you from playing hockey? Please explain.
4. Do you have any vision, hearing, or balance impairments? Have these ever prevented you from playing hockey? Please explain.
5. Is there any other reason why you believe you shouldn’t participate in this study? Please explain.

Appendix II – Subjects Measurements

Measurement Name	Left Value	Right Value	Model Filter	Measurement Description
Mass (kg)			Upper, Lower & All	Weight of subject. Unit conversion 2.2 lbs = 1 kg
Height (mm)			Lower & All	Height of subject. Unit conversion 1 in. = 2.54 cm
Inter-ASIS Distance (cm)			Lower & All	Distance between the left ASIS and right ASIS. This measurement is only needed when markers cannot be placed directly on the ASIS, for example, in obese patients.
Leg Length (mm)			Lower & All	measured between the ASIS and the medial malleolus, via the knee joint. If, for example, the subject is standing in crouch, this measurement is NOT the shortest distance between the ASIS and medial malleoli, but rather the measure of the skeletal leg length.
Knee Width (mm)			Lower & All	The medio-lateral width of the knee across the line of the knee axis. Measure this distance with the subject standing if possible.
Ankle Width (mm)			Lower & All	The medio-lateral distance across the malleoli. Measure this distance with

				the subject standing if possible.
Tibial Torsion (deg)	N/A	N/A	Lower & All	(Optional) Not necessary when using the tibial marker to identify the ankle axis.
Thigh Rotation offset (deg)	CPiG	CPiG	Lower & All	Calculated by the Plug-in-Gait model if you are using the Knee Alignment Device
Shank Rotation offset (deg)	CPiG	CPiG	Lower & All	Calculated by the Plug-in-Gait model if you are using the Knee Alignment Device
Foot Plantar Flexion offset (deg)	CPiG	CPiG	Lower & All	Calculated by the Plug-in-Gait model
Foot Rotation offset (deg)	CPiG	CPiG	Lower & All	Calculated by the Plug-in-Gait model
Shoulder Offset (mm)			Upper & All	vertical offset from the base of the acromion marker to shoulder joint center
Elbow Width (mm)			Upper & All	Width of elbow along flexion axis (roughly between the distal epicondyles of the humerus)
Wrist Width (mm)			Upper & All	Anterior/ Posterior thickness of wrist at position where wrist marker bar is attached.
Hand Thickness (mm)			Upper & All	Anterior/ Posterior thickness between the dorsum and palmar surfaces of the hand.
Shoulder width(cm)				Distance b/w the 2 Acrioclavicular joints. Take measurement from back of neck.