HAND-, ELBOW- AND SHOULDER-TO-HEAD CHECKS IN ICE HOCKEY: WHICH SCENARIO CREATES THE GREATEST PEAK LINEAR AND ROTATIONAL ACCELERATIONS OF THE HEAD?

Potvin, B.M.[1], Sidhu, A.[1], Aguiar, O.[1], Robinovitch, S.N.[1]
[1]Biomedical Physiology and Kinesiology, Simon Fraser University, Burnaby, BC, Canada
Email: potvin@sfu.ca

INTRODUCTION

Ice hockey has the highest risk for concussion among team sports in Canada[1] and 62% of concussions in elite play are due to impact to the head by the opposing player's shoulder, elbow or hand [2]. We compared head impact severity during shoulder checks, elbow checks and hand-to-head contacts.

METHODS

Participants (n=11) were players aged 21-25yrs (mean height=178.5cm (SD= 6.5); mean body mass=83.2kg (SD=6.1)) on the Simon Fraser University men's hockey team. During the experiment, participants delivered the "hardest hit they were comfortable in delivering" with the shoulder, elbow or hand, to the front or lateral aspect of a Hybrid III head with a 3-2-2-2 accelerometer array housed in a kickboxing dummy. Participants were allowed a 2 m run-up. We acquired four trials in each condition in a randomized order. A helmet was secured to the head of the dummy, and participants wore standard hockey shoulder pads, elbow pads and hands. Whole-body kinematics (8-camera Qualisys Miqus) were recorded at 640 Hz, and head accelerations were acquired at 20 kHz. Repeated measures ANOVA (JMP 13) was used to test the effect of participant impact site (hand, elbow or shoulder) and head impact site (front or lateral) on peak linear head acceleration (A_LIN), peak rotational head acceleration (A_ROT), and impact velocity (V).

RESULTS AND DISCUSSION

Participant impact site associated with A_LIN, A_ROT, and V ($p<0.0001$ for all three; Fig. 1). The mean value of A_LIN was 76% greater for hand (18.2*g* (SD=5.7)) and 27% greater for elbow (14.1*g* (SD=4.3)) than for shoulder (11.1*g* (SD=3.0)). A_ROT was 92% greater for hand (1058rad/s^2 (SD=299)) and 67% greater for elbow (919rad/s^2 (SD=372)) than shoulder (551rad/s^2 (SD=184)). V was 73% greater for hand (4.29m/s (SD=0.54)) and 108% greater for elbow (5.18m/s (SD=0.61)) than for shoulder (2.48m/s (SD=0.57)). Head impact site associated with peak A_ROT ($p=0.02$; averaging 21% greater in lateral (923rad/s^2 (SD=388)) than front (763rad/s^2 (SD=321)), but not with peak A_LIN ($p=0.8$) or V ($p=0.5$). Participants evaluated the similarity of the hits to on-ice hits with a mean score of 6.9/10.

We found that peak linear and rotational head accelerations were greater when players delivered impacts to the head using the hand and elbow than the shoulder, potentially due to impact velocity.

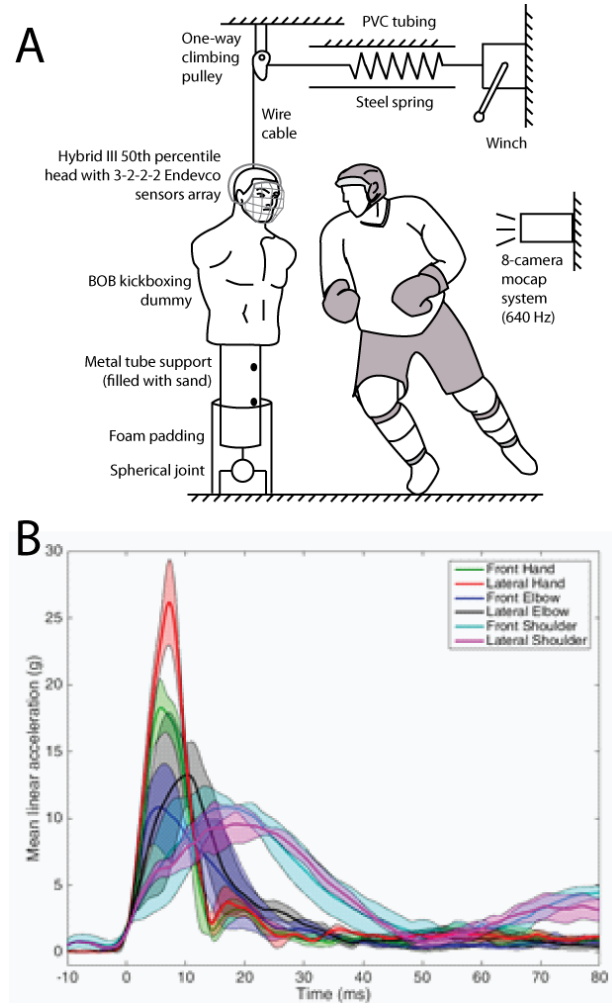

Figure 1: Experimental setup (A) and traces of mean (±SD) head form linear acceleration from one representative participant over four repeated trials the front and lateral aspect of the head by the shoulder, elbow and hand.

CONCLUSIONS

While the duration of the impact was shorter for the hand and elbow than the shoulder (Fig. 1), our results highlight the need for improvements to the design of gloves and elbow pads to reduce head impact severity in ice hockey.

REFERENCES

1.Cusimano et al., PLoS One, 2013;8(3)
2.Hutchison MG, et al. Br J Sports Med 2015, 49:552–555

ACKNOWLEDGMENTS

This study was funded in part by NSERC (RGPIN/04065).

FIRST METATARSOPHALANGEAL JOINT KINEMATICS MEASURED USING BIPLANE FLUOROSCOPY

Christopher Prasanna[1], Eric Thorhauer[1,2], Matt Kindig[1], Joseph Iaquinto[1,2], William Ledoux[1,2,3]
[1] RR&D Center for Limb Loss and MoBility, U.S. Department of Veteran Affairs, Seattle, WA USA
[2] Department of Mechanical Engineering, University of Washington, Seattle, WA USA
[3]Department of Orthopaedics & Sports Medicine, University of Washington, Seattle, WA USA
email: c.prasanna.va@gmail.com, web: https://www.amputation.research.va.gov/

INTRODUCTION

The first metatarsophalangeal joint (MTPJ1), the joint that connects the first metatarsal and first proximal phalanx of the toe, is an important part of human locomotion. It is often affected by osteoarthritis, a disease that can be treated with surgical replacement (arthroplasty). Artificial joints must be designed to mimic native behavior, however, quantifying the motion of this joint is challenging. Traditional motion capture suffers from skin motion artifacts, in which the markers (adhered to the skin or shoe) can move relative to the underlying bones of interest. Additionally, tracking the kinematics of the foot while shod is difficult, requiring either holes cut in the shoe to access the skin (changing the shoe's structural behavior) or placing markers on the shoe (which causes substantial errors).

As an alternative, our laboratory has developed a biplane fluoroscopy system to directly capture images of foot and ankle bones using X-rays. The biplane system consists of a walkway surrounded by two fluoroscopy units; high-speed video cameras capture the raw X-ray images at 500Hz. The images are then converted to 3D kinematics using a model-based bone tracking algorithm [1]. However, it is important to establish how accurately we can track this motion using this system. Therefore, in this study, we tracked MTPJ1 motion during simulated gait using cadaveric feet with beads embedded in the bones (our gold standard) to evaluate the system's performance.

METHODS

For this study, three 3mm stainless steel beads were embedded into each of the first metatarsal and first phalanx of two cadaveric feet. Each foot was mounted to a custom rig (Figure 1) consisting of a carriage that was manually pushed along a cam profile to drive the tibia in a physiologic motion in the sagittal plane. Meanwhile, cables were attached to the tibialis anterior, Achilles tendon, extensor hallucis longus, and flexor hallucis longus tendons to allow for manual muscle actuation. We collected multiple trials each for barefoot and shod conditions.

Once the imaging data was collected, processing the bead positions was done using an in-house MATLAB GUI. This program fits a Gaussian distribution to each X-ray image's grayscale intensity to define each bead's centroid; the direct linear transform (DLT) algorithm was then used to determine the 3D coordinate of each bead from the 2D image locations [2]. Once this step was done, the bead data was used to model the first metatarsal and phalanx as rigid bodies in order to obtain kinematic results.

Figure 1. A screenshot of an X-ray frame (left) of a cadaver while mounted to the custom rig (right). Beads are shown as six black circles on the first metatarsal and phalange.

RESULTS AND DISCUSSION

MTPJ1 flexion (sagittal plane) and internal/external rotation (transverse plane) were of primary interest; for brevity, only the sagittal data is reported (Figure 2). As expected, range of motion in the sagittal plane was markedly reduced in the shod condition due to the interference caused by the shoe material. Also note that the inconsistency between condition trials is due to the operation of the rig. An in vivo study would be more physiologically accurate.

CONCLUSIONS

Two cadaveric feet were manually moved within a biplane fluoroscopy system while stainless steel beads were tracked. Now that the bead data has been analyzed, the next step will be to test the model-based bone tracking method on the same data. This way direct comparison between marker and model-based tracking methods can be made in order to validate the model-based tracking's accuracy.

REFERENCES

1. Iaquinto JM, et al., *Computers in Bio & Medicine*, 2018.
2. Iaquinto JM, et al., *Medical Engineering & Physics*, 2014.

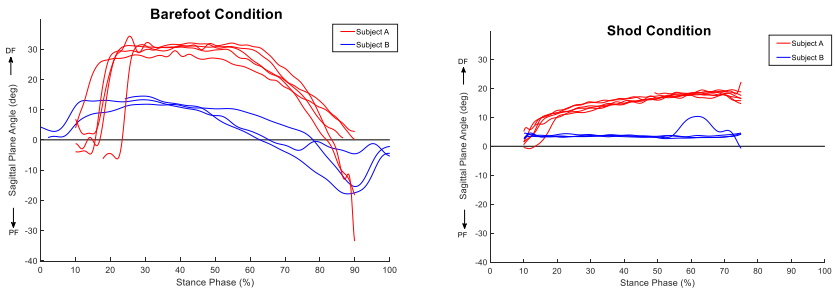

Figure 2. MTPJ motion in the sagittal plane (all trials) for barefoot (left) and shod (right) conditions.

MOTIONS OF THE HIP AND KNEE DURING SINGLE-LEG STEP-DOWN TEST AND RUNNING

Brocato, C. J.[1], Suprak, D. N.[1], Buddhadev, H. H.[1], Olinger, K. M.[1], Corl-Baietti, B. A.[1], Noble, L. K.[1], and San Juan, J. G.[1]
[1] Health and Human Development, Western Washington University, Bellingham, WA USA
email: brocatc@wwu.edu, web: https://wp.wwu.edu/biomechanicslab/

INTRODUCTION

Running is continuously growing as one of the most common methods of regular exercise in the world [1]. As the sport becomes more accessible, so does the prevalence of running-related musculoskeletal injury, with the highest overall incidence of injury occurring in the knee [2]. The source of many of these injuries is still investigated, including the prevalence of patellofemoral pain (PFP) [3]. The step-down test (SDT) is a common test used in analyzing kinematics to determine the potential source of PFP that becomes present during a run [4,5]. Similar kinematics have also been observed in individuals with PFP during a run [6], however, no known studies exist that test the relationship between the SDT and the kinematics at the midstance phase of running (MSTR). This study therefore aimed to investigate the relationship between the SDT when compared to MSTR in healthy individuals. The authors hypothesize that there is no statistically significant difference in the motion of the trunk, hip, and knee during the step-down test and mid-stance phase of running.

METHODS

Ten volunteer subjects (8 Female, 2 Male, age = 32.2 ± 7.5 years, height = 170.1 ± 6.7 cm, mass = 65.8 ± 10.0 kg, run volume = 56.3 ± 28.3 km/wk) were asked to perform two randomized conditions in a single session. Subject eligibility required that they have ≥ 6 months of regular running experience of ≥ 32 km/wk, have no history of ligament or joint reconstruction surgery to either the lower body or lower back, no history of neurological or systemic conditions that affect function of either the lower body or the lower back, and no knee pain present in either leg during any of their regular exercise or day-to-day tasks.

Subjects were instrumented with 21 reflective markers on the upper and lower extremity for motion analysis using a 10 camera VICON (Oxford, UK) set-up measuring at 250 Hz. Following a 10 minute warm-up, subjects performed either a 10 minute run on a treadmill (TR) at 3 m/s or the SDT. The SDT required subjects to tap the heel of the foot opposite of the stance leg for five times from a box adjusted to 10% of their total body height, moving at a consistent speed throughout the trial. Subjects received minimal instruction on form during the SDT but were asked to keep the heel of their stance leg flat on the box when lowering during the task.

Five running trials were recorded on the last 15 seconds of the 5th, 6th, 7th, 8th, and 9th minutes of the run. The middle 3 trials were used for analysis. The peak knee flexion (PKF) angle was found in each of the 3 trials and the joint angles of the trunk, hip, and knee were analyzed at those time points. The three time points were then averaged. This PKF angle was then used to find the time point in the SDT trials. Five step-down trials were recorded for each leg. Of the five step-down trials recorded, the middle 3 trials were averaged and analyzed at the point of the average of the PKF angles during the mid-stance phase of the run trial.

RESULTS AND DISCUSSION

A one-way repeated-measures ANOVA was used to determine the statistical significance of the joint angles of the trunk, hip, and knee between the two conditions for the dominant leg at the mean PKF during MSTR. There was a significant difference in hip abduction (HABD) ($p < 0.05$), lateral pelvic tilt (LPT) ($p < 0.05$), and trunk lateral lean (TRLE) ($p < 0.05$) between the two conditions (Figure 1).

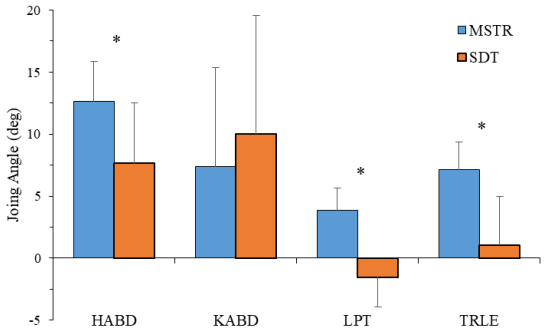

Figure 1: Frontal plane motion for dominant leg. *p < 0.05

Results of this study suggest there are kinematic differences between the SDT and MSTR. Differences observed could be due to the demand on the lower extremity during the tasks. The SDT is a slow-lowering task, meant to mimic stair descent [7,8]. The hip and knee could therefore be stabilizing the segments as the body descends, controlling and limiting the range of motion occurring in the frontal plane. Alternatively, subjects could be responding to the higher load and effect on gravity in the MSTR during a higher velocity of movement and could therefore see greater range of motion and in the joint segments during the run.

CONCLUSIONS

Further investigation should analyze the SDT at the point of PKF to observe differences in PKF of the SDT and the MSTR, and include subjects with PFP. Sex differences should also be investigated, considering there is evidence of a gender-related difference in both the SDT and running kinematics.

REFERENCES

1. Benca, E, et al., *Med Zurnal* **19**(4), 345-353, 2013.
2. van Gent, RN, et al., *Br J Sports Med* **41**, 469-480, 2007.
3. Souza, RB, et al., *J Ortho Sports Phys Ther* **40**(5), 277-285, 2010.
4. Araújo, VL, et al., *Clin Biomech* **44**, 28-35, 2017
5. Willy, RW, et al., *J Ortho Sports Phys Ther* **41**(9), 625-632, 2011
6. Noehren, B, et al., *Clin Biomech* **27**(4), 366-371, 2012.
7. Earl, JE, et al., *J Ortho Sports Phys Ther* **37**(5), 245-252, 2007.
8. Bolgla, LA, et al., *J Ortho Sports Phys Ther* **38**(1), 12-18, 2007.

ACKNOWLEDGEMENTS

We want to thank Kristian Robles for his help in collecting and analyzing the data for this study.

MODULATION OF NON-LINEAR STIFFNESS IN THE FOOT-SHOE COMPLEX ACROSS RUNNING SPEEDS

Day, E.M.[1,2] and Hahn, M.E.[1,2]
[1]Bowerman Sports Science Clinic, [2]Neuromechanics Lab, University of Oregon, Eugene, OR, USA
email: eday5@uoregon.edu, web: bssc.uoregon.edu

INTRODUCTION

The foot and shoe act in series to provide stiffness in the foot-shoe complex.[1] Increasing stiffness of the shoe increases the applied torque of the shoe during metatarsophalangeal (MTP) joint plantar flexion in late stance, but may inhibit natural motion of the MTP joint.[2] Despite the growing interest in optimally tuning footwear bending stiffness to increase running performance,[3] little is known about the individual contributions of the foot and shoe to MTP joint stiffness across running speeds. The purpose of this study is to determine the stiffness contributions and temporal behavior of the foot and shoe to MTP joint stiffness across running speeds.

METHODS

To date, eleven male runners (16:11 5000m best, 54 mi/wk) have been analyzed. Running trials were conducted on an instrumented treadmill (Bertec, Inc.) at 3.89, 4.44, 5.00, 5.56, and 6.11 m/s. Data were collected for ten strides per speed. Subjects all wore the same footwear to control for the effects of varied longitudinal bending stiffness on MTP joint mechanics. Windows were cut in the shoes at the hallux, first and fifth metatarsals, and the medial, lateral, and posterior calcaneus to allow for marker placement.

A two-segment foot model was defined. The MTP joint was modeled as a hinge axis defined by the vector between the first and fifth metatarsal markers. The moment arm for kinetic analysis was defined as the perpendicular distance from the center of pressure to the MTP joint axis. The forefoot mass was considered negligible. The MTP joint moment was considered zero until the center of pressure acted distal to the MTP joint axis.

The slow (3.89 m/s), medium (5.00 m/s) and fast (6.11 m/s) speeds were analyzed. Two measures of joint stiffness, active ($K_{active}=\Delta M_{peak}/\Delta\Theta$) and critical ($K_{cr}=\Delta M/\Delta\Theta_{peak}$) were defined. These values represent stiffness of the foot-shoe complex. Instantaneous stiffness was calculated as the first derivative of the MTP joint angular load-displacement curve. Footwear stiffness ($K_{shoe}=\Delta M/\Delta\Theta$) was calculated from mechanical testing data provided by the manufacturer. Between speed effects were analyzed using a repeated measures ANOVA with Bonferroni adjustment.

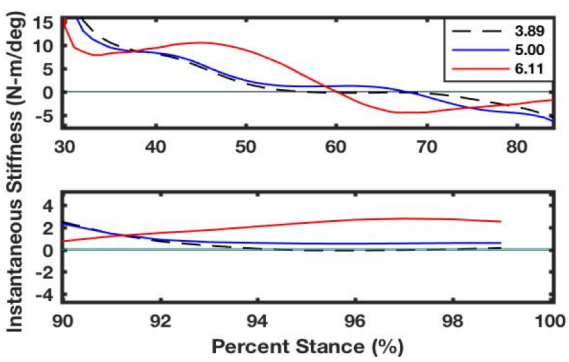

Figure 1. Instantaneous MTP joint stiffness across speeds (m/s) during loading (top) and push-off (bottom). Horizontal line represents footwear stiffness.

RESULTS AND DISCUSSION

Data presented from three of five running speeds are part of an ongoing study. K_{cr} (N-m/kg/deg) was significantly higher during fast (0.0164±0.0045) compared to slow (0.0083±0.0034) and medium (0.0121±0.0059) running speeds ($p < .005$). K_{shoe} was linear from zero to 30 degrees of flexion, 0.0727 N-m/deg. K_{active} was not different between speeds ($p = .585$).

Stiffness of the lower extremity joints are commonly analyzed using linear equations. However, the MTP joint stiffness curve elicits a non-linear pattern.[2] It may be more appropriate to utilize the derivative of the angular load-displacement curve to understand how MTP joint stiffness varies throughout stance. Instantaneous stiffness of the foot-shoe complex was much larger than the shoe and variable throughout stance, suggesting that the foot modulates foot-shoe complex stiffness (Figure 1). While the stiffness values of the shoe and foot-shoe complex were approximately equivalent from 50-65% stance at slower speeds, the MTP moment was much larger than that required to bend the shoe. The low stiffness value is representative of little change in MTP moment while MTP dorsiflexion increases.

The instantaneous stiffness curve may be utilized to understand when varying amounts of work are being done on the shoe by the foot. High stiffness represents phases when a large amount of work is being performed, whereas low stiffness represents energy dissipation. This analytical approach may be of use to understand which sections of a shoe should consist of high energy storage and return materials and which sections should facilitate a quick roll-through to minimize energy dissipation.

CONCLUSIONS

Examination of MTP joint stiffness during running suggests that the foot modulates stiffness of the foot-shoe complex and that it increases with running speed. These results may be of application in shoe or prosthetic foot design.

REFERENCES

1. Kelly et al. *J. R. Soc. Int.* **13,** 20160174, 2016
2. Oh and Park, *J. Biomech.* **53**, 127-135, 2017
3. Hoogkamer et al. *Sports Med*. **47,** 1739-1750, 2017

ACKNOWLEDGEMENTS

Thank you to Brooks Sports for donating footwear for this study.

The Influence of Shoe Type on Running Biomechanics: A Comparison of Minimal, Maximal, and Traditional Running Shoes.

Diulio, J.F., Hannigan, J.J., Ter Har, J.A., and Pollard, C.D.
FORCE Laboratory, Kinesiology Program, Oregon State University-Cascades, Bend, OR USA
email: christine.pollard@osucascades.edu, web: http://osucascades.edu/force-lab

INTRODUCTION

Running footwear aims to reduce high impact forces during running; however, there have been no significant changes in injury rates of recreational runners over the last 50 years [1]. While numerous studies have investigated the effect of minimal running shoes on running biomechanics and injury rates [2], very little research has been performed on maximal cushioned shoes, such as the Hoka One One, which claims to reduce the impact forces of running. Furthermore, no study has concurrently studied the effect of minimal, traditional, and maximal running shoes on running biomechanics. Therefore, the purpose of this study was to compare ground reaction forces and ankle kinematics between a minimal, traditional, and maximal running shoe in healthy, recreational runners.

METHODS

Eight male and nineteen female recreational runners (ages 18-45) who ran a minimum of 10 miles a week participated in this study. Each subject had no prior training in either minimal or maximal running shoes 6 months prior to the study. All participants were rear foot strikers in all shoe conditions.

Subjects participated in one data collection where they ran in a minimal, traditional, and maximal New Balance running shoe (order of footwear randomized) (Figure 1). Subjects ran along a 10-meter runway while 3-D kinematics were collected using an 8-camera motion-capture system collecting at 250 Hz (Vicon Motion Systems, Oxford UK). Two embedded force plates collected ground reaction forces at 1000 Hz (AMTI, Watertown MA). Five successful running trials for each shoe condition.

Data was processed with Vicon Nexus software while kinematics and ground reaction force data were calculated using Visual3D software (C-Motion, Germantown MD). The variables of interest included impact peak, overall peak, and loading rate of the vertical ground reaction force, and ankle kinematics for dorsiflexion and eversion (angle at initial contact, peak angle, angle at toe-off, and joint excursion). Data were analyzed using a repeated measures ANOVA with the alpha-level set to 0.05.

RESULTS AND DISCUSSION

There were no significant differences between the three shoe conditions for impact peak, overall peak, and loading rate of the vertical ground reaction force.

Subjects exhibited greater ankle dorsiflexion at initial contact in the traditional shoe compared to the minimal shoe (traditional: 9.9 ±3.4°; minimal: 3.6 ± 9.7°; p = 0.009), as well as greater ankle dorsiflexion excursion in the minimal shoe compared to both traditional and maximal shoes (traditional: 13.2 ± 3.2°; maximal: 13.6 ± 3.4°; minimal: 19.2 ± 8.3°; p = 0.003).

Peak ankle eversion was significantly greater in the maximal shoe compared to traditional (traditional: -10.6 ± 3.6°; maximal: -12.1 ± 3.6° BWs p = 0.007). The ankle was also significantly more everted at toe-off in the maximal shoe compared to traditional and minimal (traditional: 0.56 ± 4.4°; maximal: -1.1 ± 4.5°; minimal: 1.7 ± 4.6°; p = 0.014). There was a trending but non-significant difference in ankle eversion excursion, with the maximal shoe greater than the traditional (traditional: -11.6 ± 4.5°, maximal: -12.4 ± 4.6°, p = 0.100).

Figure 1. From left to right, the minimal, traditional, and maximal running shoes utilized in this investigation.

CONCLUSIONS

The findings for the vertical ground reaction force data contradict previous findings that running in a maximal shoe increases the impact peak and loading rate compared to a traditional shoe [3]. We believe this difference may be due to differences in shoe construction between the New Balance maximal shoe utilized in this study and the maximal Hoka One One shoe used in the previous study.

In the maximal shoe, subjects ran exhibited greater peak eversion and were still everted at toe-off. This finding may indicate that running in a maximal shoe leads to prolonged eversion across the stance phase, which may increase the risk of developing Achilles tendinopathy and medial tibial stress syndrome [4].

A limitation of this study is that each shoe was novel for the subjects. Future studies should allow the subjects to get accustomed to the different types of footwear.

REFERENCES

1. Taunton JE, et al. *Br J Sports Med* **36**, 95-101, 2002.
2. Sinclair J. *Clin Biomech* **29**, 395-9, 2014.
3. Pollard CD, et al. *Orth J Sports Med* (in press). 2018.
4. Becker J, et al. *Am J Sports Med* **45**, 2614-21, 2017.

SEX IMPACTS LEG STIFFNESS WHEN INCREASING STRIDE LENGTH TO RUN WITH BODY BORNE LOAD

[1] Nicholas J. Lobb, [1]AuraLea C. Fain, [1]Kayla D. Seymore, and [1]Tyler N. Brown
[1] Boise State University, Boise, ID, USA
email: nicklobb@u.boisestate.edu

INTRODUCTION

During military training, personnel are required to run at a fixed cadence with heavy body borne loads. These loads routinely weigh between 20 kg and 40 kg, and alter lower limb biomechanics [1]. When running with load, personnel increase leg stiffness to attenuate larger GRFs and prevent collapse of the lower limb, increasing risk of musculoskeletal injury [2]. When running without load, individuals decrease leg stiffness by increasing lower limb flexion as stride lengthens [3]. Military personnel, however, may not possess the lower limb strength to decrease leg stiffness with the longer strides required to run at a fixed cadence with heavy body borne loads. Female personnel may be especially at risk as they are purportedly weaker than males [4]. This study sought to quantify leg stiffness for males and females as they lengthened their stride to run with body borne load.

METHODS

Seventeen male and ten female participants (21.2 ± 2.3 yrs, 1.7 ± 0.1 m, 75.5 ± 11.3 kg) had 3D lower limb biomechanics quantified while running with four load conditions: 20 kg, 25 kg, 30 kg, and 35 kg. For each load, participants wore a helmet, weighted vest, and carried a mock weapon. The vest weight was systematically adjusted to apply the load necessary for each condition. For the run task, participants were required to run at 4 m/s ± 5% using one of three stride lengths: preferred stride length (PSL), 15% shorter than PSL (SSL), and 15% longer than PSL (LSL). Participants performed five successful trials at each stride length, which required they run with the correct stride length and speed, and only contact the force platform with their dominant leg.

During each run trial, lower limb biomechanics were quantified from the 3D trajectories of 34 reflective markers. Synchronous GRF data and marker trajectories were low pass filtered with a fourth-order Butterworth filter (12 Hz). Then, filtered marker trajectories were processed using Visual 3D (C-Motion, Rockville, MD) to solve joint rotations at each instant. Leg stiffness was calculated as the GRF vector directed through the hip joint center divided by the maximum change in leg length [5].

$$\text{Leg Stiffness: } K_l = \frac{F_e}{\max(\Delta L_e)}$$

For analysis, leg stiffness, and peak of stance (PS, 0%-100%) vertical GRF (vGRF), change in leg length, and hip, knee, and ankle flexion angles were quantified. Each variable was submitted to a RM ANOVA to test the main effect and interaction between load (*20 kg, 25 kg, 30 kg, 35 kg*), stride length (*SSL, PSL, LSL*), and sex (*Male, Female*). Significant interactions were submitted to a simple effects analysis, and a Bonferroni correction was used for multiple comparisons. Alpha was $p < 0.05$.

RESULTS AND DISCUSSION

Running with body borne load increased risk of injury. Participants exhibited a significant increase in leg stiffness (P=0.006) and peak vGRFs with the addition of load (P<0.001) (Table 1). The stiffer leg may increase transmission of the vGRF to the lower limb, increasing risk for injury. This risk may further increase when running with longer strides. Peak vGRF (P<0.001) increased with stride length. Males, however, decreased leg stiffness when using LSL compared to PSL (P<0.001) and PSL compared to SSL (P=0.026); whereas, females exhibited no difference in leg stiffness between strides (P>0.05). The discrepancy in leg stiffness may be attributed to a sex dimorphism in lower limb biomechanics. Although, females adopted greater PS hip and knee flexion compared to males with SSL (P=0.013; P=0.001), only male participants may possess the strength to significantly increase PS hip and knee flexion as stride length increased from SSL to PSL (P=0.008; P=0.001) and from PSL to LSL (P=0.041; P<0.001). Further, participants increased PS knee flexion with LSL compared to PSL when carrying 20 kg (P=0.001), 25 kg (P<0.001), and 30 kg (P=0.004), but not 35 kg (P=0.760). These heavy body borne loads may require greater lower limb strength to increase lower limb flexion and attenuate the larger vGRF to reduce the elevated injury risk of running with longer strides.

CONCLUSIONS

Body borne load increased leg stiffness and potential risk of musculoskeletal injury during running. Increased leg stiffness may help attenuate the greater peak vGRFs and prevent collapse of the lower limb when running with heavy body borne loads. Only male participants, however, demonstrated the ability to reduce leg stiffness and potentially risk of injury by using greater lower limb flexion when running with longer strides. Further study is warranted to determine if lower limb strength, rather than sex, determines the ability to adjust leg stiffness and lower limb biomechanics when altering stride length to run with heavy body borne load.

REFERENCES

1. Patton JF, et al. *Eur J Appl Physiol*, **63**:89-93, 1991.
2. Silder A, et al. *J Biomech*, **48**:1003-1008, 2015.
3. Farley CT, et al. *Clin. Biomech*, **26**: 23-28, 1998.
4. Miller A, et al. *Eur J Appl Physiol*, **66**: 254-262, 1993.
5. Coleman D, et al. *J Biomech*, **45**: 1987-1991, 2012.

Table 1: Mean (SD) Leg Stiffness and Peak vGRF Exhibited by Males and Females with Changes in Stride Length and Load.

		20 kg		25 kg		30 kg		35 kg	
		Male	**Female**	**Male**	**Female**	**Male**	**Female**	**Male**	**Female**
Leg Stiffness	**SSL**	30.1 (7.9)	26.3 (6.8)	32.9 (8.7)	27.5 (6.9)	33.7 (9.2)	27.5 (5.1)	33.1 (7.4)	28.2 (5.4)
	PSL	28.0 (5.9)	25.6 (4.4)	28.9 (7.0)	28.8 (8.2)	30.8 (6.6)	28.0 (7.5)	32.8 (7.1)	28.8 (5.6)
	LSL	24.0 (5.9)	22.8 (4.6)	25.7 (6.3)	25.5 (6.4)	25.6 (5.6)	26.3 (7.1)	27.0 (6.3)	26.3 (6.2)

A RELIABLE AND CLINICALLY FEASIBLE IMU-BASED DUAL-TASK GAIT BALANCE CONTROL ASSESSMENT

Pitt, W.J.[1] and Chou, L.S.[1]
[1]Departmnet of Human Physiology, University of Oregon, Eugene, OR, USA
email: chou@uoregon.edu, web: http://choulab.uoregon.edu

INTRODUCTION

Concussion in athletics, workplace, and everyday activities has been shown to lead to impaired dynamic balance control persisting as long as two months post-injury [1]. As current concussion assessments are unable to detect these impairments, individuals may return to activity prior to complete recovery, resulting in increased risk for repeated brain injury, musculoskeletal injury, and the development of chronic symptoms. At present a clinically feasible dual-task gait balance assessment utilizing sensitive whole body center of mass (COM) biomechanical markers is not available. The purposes of this study are therefore to examine the internal consistency and interrater reliability of a novel IMU-based dual-task gait balance control assessment and to demonstrate its clinical feasibility with application in Division 1 athletics.

METHODS

In the first part of this two-part study, 20 (10F) subjects performed four walking trials in each of three conditions: walking, walking while performing an auditory Stroop test, and walking while spelling a five letter word backwards or subtracting from a given number by 6s or 7s. Individuals walked at a self-selected pace over an eight meter level path, performed a 180° turn, and returned to the start. Conditions were repeated in two environments (lab and hallway similar to a medical clinic), by two raters, and on two days separated by 7-10 days. The order of environment, rater, and condition was randomized for each testing session. An eight item Cronbach's α was used to establish the consistency of the tool across time, environment, and rater. An ICC and 95% CI was used to establish interrater reliability.

Instrumentation consisted of an APDM OPAL IMU system (sensor placed over the L5 vertebrae as a COM proxy [2]) for kinematic data collection, Bluetooth headset/microphone for assessment instruction and cognitive test application, and Superlab 5 software for protocol automation. Hardware and software were controlled with a single MacBook Air laptop. IMU data was sampled at 128Hz and filtered with a 2nd order, low-pass, Butterworth filter, with a 12Hz cutoff. The gait cycles beginning with the 5th heel strike during the walk out and the returning walk were analyzed. Data were normalized to a single gait cycle with a left foot heel strike initiation. Eight peak linear accelerations along the vertical, medial-lateral, and anterior-posterior axes were identified (Fig. 1).

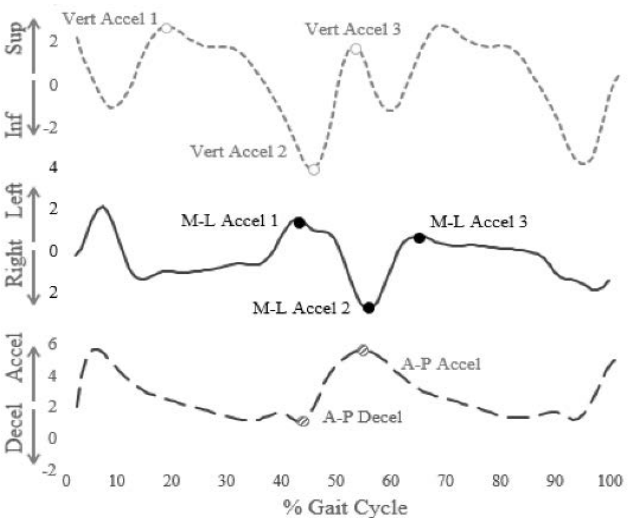

Figure 1: Representative plot of the Vert, M-L, and A-P linear accelerations and 8 peak accelerations (m/s^2)

The second part of the study consisted of a shortened clinical protocol consisting of three walking trials in each of the three randomly presented conditions, implemented with 14 athletes from the women's soccer team. Total assessment time per athlete was collected to assess clinical feasibility. Additionally, the dual-task cost (DTC) of the eight linear acceleration metrics were collected and compared to non-athlete females from part one of the study using a Multivariate Analysis of Variance (MANOVA) for each of the two dual-task conditions separately.

RESULTS AND DISCUSSION

34 (24F) total subjects completed the study. Cronbach's α values for all eight metrics in the three walking conditions had a range of .881 to .980 indicating high to very high internal consistency. Interrater reliability was excellent as demonstrated by ICC values of .868 to .983 (table 1). Total assessment time for female athletes including sensor placement and verbal instructions was 8:30 min ±35 sec. Wilk's test of multivariate significance was not significant, p<.713 and p<.256, for DTC between walking to Stroop and walking to Q&A respectively, suggesting membership in different groups of females were not related to the weighted multivariate combination of peak acceleration DTCs.

Metric	Cronbach's α			ICC	
	Walk	Stroop	QA	r	95% CI
Vert Accel 1	.976	.942	.976	.920	.918±.022
Vert Accel 2	.951	.955	.980	.868	.865±.033
Vert Accel 3	.911	.914	.933	.876	.872±.034
M-L Accel 1	.931	.963	.969	.931	.928±.020
M-L Accel 2	.933	.904	.925	.902	.898±.027
M-L Accel 3	.976	.969	.970	.941	.939±.016
A-P Decel	.906	.889	.881	.983	.981±.006
A-P Accel	.918	.930	.925	.983	.982±.005

Table 1: Cronbach's α and ICC values for the 8 metrics

CONCLUSIONS

An IMU-based dual-task gait balance control assessment using COM kinematic measures demonstrates high internal consistency between environment, time, and rater, and has excellent interrater reliability. Furthermore, the instrument may be feasibly applied in clinical settings.

REFERENCES

1. Howell DR, et al. *Arch Phys Med Rehabil.*, **94**(8), 513-1520, 2013
2. Howell DR, et al. *JOB.*, **48**(23), 3364-8, 2015